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Optimization
of
Chest Radiography

Proceedings of a Symposium
held in
Madison, Wisconsin
April 30 — May 2, 1979

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Optimization
of
Chest Radiography

Proceedings of a Symposium
held at
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Madison, Wisconsin
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Principal Investigators;
and John F. Wochos, Program Coordinator
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WHO Collaborating Center for
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FOREWORD

The Bureau of Radiological Health develops and carries out a national program to control unnecessary human exposure to potentially hazardous ionizing and nonionizing radiations and to ensure the safe, efficacious use of such radiations. The Bureau publishes the results of its work in scientific journals and in its own technical reports.

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The Bureau also makes its technical reports available to the World Health Organization. Under a memorandum of agreement between WHO and the Department of Health and Human Services, three WHO Collaborating Centers have been established within the Bureau of Radiological Health, FDA:

> WHO Collaborating Center for Standardization of Protection Against Nonionizing Radiations;
>
> WHO Collaborating Center for Training and General Tasks in Radiation Medicine; and
>
> WHO Collaborating Center for Nuclear Medicine.

Please report errors or omissions to the Bureau. Your comments and requests for further information are also encouraged.

John C. Villforth
Director
Bureau of Radiological Health

PREFACE

Increased concern by the medical profession, government agencies, and the public regarding radiation exposure in general has focused attention on diagnostic radiology. X-ray examinations of the chest, although relatively low in exposure, accounted for approximately 46 percent of all medical x-ray examinations reported in the 1970 X-Ray Exposure Study conducted by the Public Health Service. Because so many chest examinations are performed annually, the potential health benefits from improvements in equipment and techniques for these examinations are very substantial.

This Symposium was intended to provide a forum for the exchange of information on the relationship between patient exposure and image quality in chest radiography. It originated from discussions of Task Group 13 of the Centers for Radiological Physics. Topics identified by the Task Group were of direct program interest to the Bureau of Radiological Health's Division of Training and Medical Applications. The University of Wisconsin was selected to organize and conduct the Symposium under a contract with the Bureau.

These Proceedings include the technical papers and discussions presented at the Symposium, conclusions and recommendations that evolved from several workshops at the Symposium, and a bibliography prepared by the University of Wisconsin as a background document for participants. The Proceedings should provide a basic reference to the state-of-the-art in chest radiography and an indication of areas in which future work is needed.

Wm S Properzio

William S. Properzio, Ph.D.
Director
Division of Training and
Medical Applications

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INTRODUCTION TO SYMPOSIUM

Gerard T. Scanlon, M.D.
Medical College of Wisconsin

As John Cameron said, I'm from the Medical College of Wisconsin, which is the other Wisconsin school in Milwaukee. I was pleased that Dr. Cameron asked me here in the spirit of ecumenism. I welcome you all to Wisconsin.

When I first heard about this conference, I was extremely pleased with the whole concept. Because of my own interest in chest radiology, I felt the Conference was needed. As I looked over the detailed plans, I had a second reaction. Was this overkill? We have convened around 40 experts from all over the country to talk for three days on what amounts to one radiological examination. That sounds like overkill when you first think about it. I thought about it further, however, and had to conclude that it was not overkill. I think the chest x ray , as all of us recognize, can epitomize all those things we study in radiology. It requires all the skills of radiologic technology to control film/screen combinations, processing, and very subtle degrees of density and contrast. Likewise, fog, scatter radiation, and radiation to the patient must be carefully controlled. So, there is a problem. How big is the problem? I would say, pretty big.

When I asked my colleagues and friends at the American College of Radiology how big the problem is, they told me over 65 million examinations of the chest were performed in the last year for which they have any numbers. Sixty five million! Now some of these were photofluorograms, but there were still almost 50 million chest x rays done, including about 35 million in hospitals alone. Fifty million chest x rays! As the radiologists in this room certainly know, about 40 percent of all our studies are examinations of the chest. What we may not know is that, stated another way, almost 50 percent of our patients get chest x rays. One of every two patients entering our department is there to get a chest x ray. If we take a very conservative $20 per examination, we are talking about over $1 billion a year for chest x ray examinations. Over $1 billion! I submit that this is not an examination--it is a virtual industry--and a billion dollar one at that. As an industry, or as a medical examination, it needs critical examination, and that is what we are all here for. We are here to examine all aspects. First of all is the financial, which we have already looked at to some degree, because it impacts on all our patients. Can we do it cheaper, or can we get more "bang for the buck" as the saying goes.

How about the medical aspects? Are we doing the exams in such a way that we can derive all possible, accurate clinical information? The answer is probably "no," but we are all here to examine why not and to consider how to improve. Last, but certainly not least, is the radiologic exposure. The radiation exposure to any given patient is quite small, and we are reducing it all the time through such measures as improved film/screen combinations. Still, it is an impact upon the entire population when we do 50 million examinations a year. That is a lot of radiation to the American public.

Next, we should consider whether the chest x ray is the exam we should be doing in all cases. I am not sure of the answer. Except for high kV techniques, which have come along in the last few years, the standard, basic chest x ray has not changed in perhaps decades. With newer technology, which some of us are going to talk about in the next two or three days, perhaps we will find there is a different type of examination we can do which will be beneficial in all the areas I mentioned--financial, clinical, and radiation exposure. I would advise a bit of caution, however, in moving into different types of examinations. The chest x ray is a pretty good radiological exam. In fact, it is very good. We derive an enormous amount of information from it; regardless of what it costs, it is still probably a bargain. As

evidence of this, look at the effect of computed tomography. When we consider the impact of computed tomography on radiology, we realize the place it has impacted the least is the chest. Why is this so? Because chest x rays are so good, we can derive an enormous amount of information from them; we do not really need computed tomography.

This Conference, then, was called to addess these questions. Dr. Cameron has assembled groups of experts to examine all questions in these fields; to tell us how to make the exam better, perhaps to tell us how to interpret better, and finally to explore with us the newer methods that may some day complement or even replace the exam we are now using.

DISEASE SIGNS IN CHEST RADIOGRAPHY

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The single most important decision the radiologist makes when interpreting a radiograph is whether the examination is normal or abnormal. His ability to accurately make this distinction depends upon a number of factors. Perhaps most important is extensive clinical experience, acquired through years of study, that enables the radiologist to distinguish the wide range of normal variations from true abnormalities. He must also, of course, be able to recognize a variety of abnormalities that reflect underlying disease processes. Inasmuch as these abnormalities are merely abnormal variations of normal anatomy, familiarity with normal radiographic anatomy again becomes a critical factor in the interpretation of radiographs. The radiologist's abilitiy to recognize both normal and abnormal anatomy is predicated on the fact that the radiographic technique accurately portrays the underlying gross anatomy. As is often said, it is not too difficult to be a good radiologist if the films are of good quality--that is, they faithfully demonstrate the underlying normal anatomy as well as any pathologic alterations that may be present.

In all radiography the underlying anatomy is visualized because of contrast between structures of differing radiographic densities. In the chest these densities are the equivalent of those seen with air, fat, water, and bone. It is often difficult, however, to distinguish between fat and water densities on standard chest radiographs; so, in essence, only three general categories of radiographic density are dealt with. In most instances in chest radiography, air-containing structures (lungs, major airways including trachea and bronchi, and esophagus) are contrasted against soft-tissue-density structures (mediastinum, heart, great vessels, and pleural surfaces). In addition, the blood vessels (both arteries and veins) coursing throughout the lung fields are contrasted against the surrounding air-containing pulmonary parenchyma. When films are analyzed for normality or abnormality, as well as evaluated for technical quality, it is generally these normal structures that are looked at; for in the end it is the radiologist's decision as to whether the structures visualized are normal or abnormal that will prove critical.

Inasmuch as familiarity with normal anatomy is the basis for interpretation of chest radiographs, a brief review of pertinent anatomic features is offered. For the purpose of this presentation, only the normal anatomy of the mediastinum and lungs will be considered. The osseous and soft-tissue structures, although of great practical importance in actual clinical interpretation of radiographs, will not be included in the present discussion.

MEDIASTINUM

Many of the most important anatomic features present in the thorax occur within the mediastinum, as the soft-tissue structures in this area are contrasted against adjacent air-containing structures. Alterations in the normal interfaces thus formed have proven to be sensitive indicators of intrathoracic disease. Familiarity with these structures, therefore, is critical to accurate interpretation of chest radiographs.

JUNCTION LINES

Where the two lungs closely abut one another with thinned mediastinum between them, a so-called "junction line" is formed. The anterior junction line, located in the retrosternal area, anterior to the ascending aorta and heart, is created as the antero-medial portions of the right and left lungs approach one another in the mid-line. The line, visualized in 20 to 25 percent of normal adult PA chest films, represents the very thin tissue of the anterior mediastinum plus the adjacent parietal and visceral pleural covering of the right and left lungs. It generally originates at the level of the sterno-manubrial junction and extends downward and slightly to the left for several centimeters.

Although absence of this junction line is not a reliable sign of abnormality, clear demonstration of a normal line is strong evidence against any significant anterior mediastinal mass. A number of disease processes such as lymphadenopathy, tumor, hemorrhage, or a prominent ascending aorta can thicken or obliterate the line. Its position and orientation may also be changed by herniation of one lung across the mid-line into the opposite hemithorax.

Posterior to the trachea and anterior to the spine, the two lungs again meet near the mid-line, above the level of the aortic arch, to form the posterior junction line. The line, as reflected on the radiograph, generally projects vertically through the tracheal air column on the PA view. Unlike the anterior junction line, which is usually very thin, the posterior junction line has a wide range of normal thicknesses. In a given patient, however, progressive increase in width of the line may be due to infection, hemorrhage, esophageal dilatation, lymphadenopathy, or neoplasm. Again, herniation of either lung through the posterior-superior portion of the mediastinum will cause bowing or displacement of this line.

An interesting variation in the appearance of the posterior junction line occurs when the upper esophagus contains a small amount of air, as it often does. The "esophageal-pleural stripe" can then be identified on the radiograph as a soft-tissue line formed between air in the right lung and air contained within the esophageal lumen. It is important to recognize this line as a normal variant, and not to confuse it with a pathologic alteration.

RIGHT PARATRACHEAL STRIPE

In over half of normal adult PA chest films, a soft-tissue stripe will be seen between the aerated right lung and the air contained within the tracheal lumen. This stripe, referred to as the right paratracheal stripe, generally has a thickness of between 1 and 4 mm. Widening of the stripe above this limit is a fairly reliable sign of disease and may reflect pathologic alterations in either the mediastinum, trachea, or pleura, all of which contribute to the soft-tissue density of the stripe.

AZYGOUS VEIN

At the termination of the right paratracheal stripe, an oval density can be seen in over half of normal adult PA chest radiographs. This density represents the arch of the azygous vein as it crosses the right lateral aspect of the mediastinum at approximately the T4 or T5 level to empty into the back wall of the superior vena cava. On an upright PA chest film, the arch generally does not exceed 7 mm in width. Enlargement of the vein occurs with systemic venous congestion or when the vein acts as a collateral pathway for blood flow, as in superior or inferior vena cava obstruction and various congenital venous anomalies. Lymphadenopathy occurring in the azygous nodes may mimic a prominent azygous vein but can be differentiated by failure to change size with changes in position or intrathoracic pressure.

AZYGO-ESOPHAGEAL RECESS

Below the level of the azygous vein the right lung again invaginates deeply into the mediastinum, behind the heart and anterior to the spine. This interface of the right lung with the mediastinum forms the so-called "azygo-esophageal recess." The lung-mediastinal interface created by this recess usually follows the esophagus quite closely. Loss of the normal contours can be expected with esophageal dilatation or tumor, hiatal hernia, cardiomegaly, and subcarinal lymphadenopathy.

AORTIC-PULMONARY WINDOW

The left lung also abuts the mediastinum creating several soft-tissue/air interfaces of anatomic importance. Invagination of the lung between the aortic arch superiorly and the pulmonary artery inferiorly forms the so-called "aortic-pulmonary window." In this small space are contained the ligamentum arteriosum, the recurrent laryngeal nerve, and the ductus lymph nodes. The mediastinal pleura, reflecting from the pulmonary artery to the aortic arch, forms the lateral border of this enclosed space. The contour of this pleural reflection has proven to be a sensitive indicator of pathology in the area. Normally, the pleural reflection over the window is concave laterally; a convex bulge, particularly if developing on serial films, is most often indicative of lymphadenopathy, although other masses in the aortic pulmonary window can cause a similar appearance.

The left lung also presses against the mediastinum in its inferior portions to outline the descending thoracic aorta. Alterations in the normal contour of this interface generally reflect abnormalities of the aorta or adjacent pleura.

PARASPINAL LINES

The most posterior interface between lung and mediastinum seen on the PA chest radiograph is the paraspinal line, representing the interface between the lung and the paraspinal soft tissues. The line is not seen as frequently on the right side because the interface in this area is frequently oriented obliquely to the x-ray beam over its entire course. On the left side, however, the presence of the descending aorta requires the mediastinal pleura to turn more laterally as it reflects off the spine and a tangential interface is often created. As a result, the left paraspinal line is identified with greater frequency.

Separation of the left paraspinal line from the vertebra may be a reflection of traction by a dilated or tortuous descending aorta. Localized displacement on either side, however, may be secondary to osteophytes (somewhat more common on the right side), lymphadenopathy, neurogenic tumor, vertebral abscess (pyogenic or tuberculous), hemorrhage following aortic rupture, vertebral fracture, or vertebral involvement by tumor.

PULMONARY PARENCHYMA

The other major area of interest on chest radiographs is the pulmonary parenchyma. Here, one primarily evaluates the distribution of normal bronchovascular structures (it is primarily the vascular structures that can be seen, contrasted against the air-containing lung parenchyma that surrounds them) as well as looking for abnormal shadows cast by infiltrates or masses within the lung.

PULMONARY VASCULARITY

In the normal patient, distribution of pulmonary vascularity is very much dependent upon the position of the patient (supine versus erect) and the degree of inflation of the lungs. Normally, there is increasing pulmonary blood flow per unit volume of lung from apex to base (assuming the film is exposed while the patient is erect) and this distribution is reflected by increasing vessel size from apex to base. Alterations in this normal vascular pattern reflect abnormalities in pulmonary vascular pressure, pulmonary blood volume, or in the state of the underlying vascular bed.

Such alterations may be the first clue to clinically unsuspected disease. Therefore, accurate radiographic portrayal of vessels throughout the lung fields is of critical importance in chest radiography. Margins of the vessels must also be accurately rendered because early interstitial pulmonary edema, which is not clinically evident, will be reflected first on the radiograph as indistinctness of vessel margins. The ability to identify this change, prior to it becoming clinically evident, will allow appropriate therapeutic intervention prior to the onset of florid pulmonary edema.

The overall radiographic density of the lung fields reflects the contributions of the soft tissues, pleural space, and pulmonary parenchyma. Accordingly, alterations in density may reflect an abnormality in any of these areas. If the soft tissues and pleura are normal, however, the change in density must reflect some pathologic alteration in the pulmonary parenchyma. Such abnormalities are of either increased or decreased density relative to normal. A decrease in density, in essence, means a decrease in blood flow through the affected area of the lung. Often the disease processes that cause such alteration can be confirmed by demonstrating localized air trapping. Chest radiographs routinely are exposed at full lung capacity, but, if air trapping is suspected, films may be obtained in varying degrees of expiration. Films exposed 1 second after beginning a forced expiration have proven valuable in detecting areas of localized air trapping and may be particularly striking if the abnormality is at the level of the main or segmental bronchi. Bronchiolar abnormalities such as seen in the Sywer-James Syndrome are also well demonstrated by this type of examination.

MASSES AND/OR NODULES

Abnormal pulmonary shadows have long been one of the major areas of interest in chest radiography. Because 25 to 50 percent of the pulmonary parenchyma is obscured by overlying ribs, or by hilar and cardiac shadows, it is important that optimal radiographic techniques allow one to "see through" these areas in order to evaluate the underlying lung parenchyma. In addition, it is well known that abnormal shadows are more commonly missed in specific areas of the chest such as the perihilar, apical, retrocardiac and subdiaphragmatic regions. In part, this difficulty is because of an increased number of structures normally portrayed in the area (contributing to an increase in the "background surround"). This superimposition of normal structures over abnormal structures renders the latter less visible. Reduction in superimposition and/or background surround can be achieved by tomography, or by changing the angle of the film (lordotics, obliques, and so forth) so as to remove the overlying structures. With regard to nodular or mass densities, identification of the abnormality is probably the most important radiographic task. The radiographic demonstration of calcification (if of specific types) can allow a definitive diagnosis of a benign lesion; but, for the most part, benign and malignant lesions cannot be reliably separated on the basis of the radiographic characteristics of the mass or the mass/lung interface.

CONCLUSION

Without question, the single most important decision rendered in interpretation of chest radiographs is whether the examination is normal or abnormal. The radiologist arrives at this decision by careful analysis of well-exposed chest radiographs to which he applies his powers as an observer and the broad base of clinical experience he has gained from years of study. Radiographic abnormalities reflecting a variety of disease processes are well discussed in the standard textbooks. In essence, however, these abnormalities represent abnormal variations of normal anatomy. For this reason, radiologists, particularly when analyzing the "quality" of films, generally look at points of normal anatomy that they expect to be able to observe, and judge the quality of the film on the basis of how well these anatomic features, particularly in the lung and mediastinum, are portrayed. One of the major problems in chest radiography continues to be difficulty in adequately portraying normal anatomic features of both mediastinum and lung parenchyma on the same film. With the current receptor systems, it is often difficult to increase the density sufficiently within the mediastinum to allow clear delineation of normal anatomic features without "blackening" the lung fields too much. Conversely, radiographs exposed to "optimally" demonstrate the lung fields often leave the mediastinum underexposed. Methods by which the normal and pathologic anatomy of the chest may be more accurately portrayed on the chest radiograph, and the factors that affect the radiologist's ability to correctly identify the information presented, will be the primary focus of this conference.

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INTRODUCTION TO OBSERVER PERFORMANCE TESTS

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INTRODUCTION

The fundamental relationships between the physical properties of a medical image (such as resolution, contrast, and statistical fluctuations) and the ability of a human observer to properly detect and interpret diagnostically relevant image features are poorly understood. Most medical imaging modalities provide a broad set of possible compromises among spatial resolution, contrast, magnitude and texture of image "noise," and required patient radiation exposure, and the best compromise usually is not apparent, even in idealized situations. Further, in real diagnostic tasks, the relationships between imaging physics and observer performance are undoubtedly complicated by problems of complex background structure, normal anatomical variations, and observer training. Thus, at present, one cannot confidently predict the diagnostic performance of a medical imaging procedure from knowledge of its physical characteristics. Physical and psychophysical theories can suggest directions that one should take in attempting to optimize the performance of diagnostic imaging systems, but the inadequate bridge that exists at present between imaging physics and psychophysics requires that we finally determine experimentally what we want to know: the diagnostic decision performance that can be achieved by human observers who view images made with real medical imaging systems.

To be useful for the evaluation of medical imaging systems, a measure of human observer performance must meet several criteria: (1) it must be objective, in the sense that it is related to some standard of "truth;" (2) it must be quantitative, in order to allow conclusions to be drawn, for example, regarding the amount of additional decision performance that can be gained at the expense of some increased radiation exposure; (3) the measure must be meaningful, in the sense that two systems providing equal performance measures can be confidently interpreted as of equal benefit in a specified diagnostic task; and (4) it must be reproducible, in a statistical sense, at least within a group of observers of similar "ability." Also, the usefulness of a diagnostic performance measure is increased if it can be used, with some confidence, to interpolate or extrapolate from one diagnostic decision-making situation to another—for example, if the measure is independent of disease prevalence, or if it can be used to predict performance in some other diagnostic task. Finally, the diagnostic performance measure ideally should be usable in a cost/benefit analysis or a decision analysis of the diagnostic problem, in order to account--at least conceptually--for the undeniable fact that different types of correct and incorrect diagnostic decisions can have different utilities, which will depend on the context of the diagnostic problem, and in order to explore the optimization of diagnostic strategies.

Various measures of the diagnostic performance provided by medical imaging systems have been proposed and used in the past. The oldest measure, subjective judgment of "picture quality," is totally inadequate; it satisfies none of the criteria mentioned above. A still-common measure of performance, diagnostic "accuracy," or the fraction of all diagnostic decisions that are correct, is objective and quantitative but meets none of the other criteria. "Accuracy" depends strongly on the prevalence of disease; when 5 percent of patients have the disease, 95 percent accuracy can be obtained by blindly calling all patients negative. Also, "accuracy" does not differentiate between the two types of diagnostic error (i.e., false negative and false positive), the consequences of which may entail very different costs. Thus, two diagnostic procedures providing equal fractions of correct diagnoses may be of very different diagnostic utility (1).

More meaningfully, diagnostic decision performance can be measured by a pair of indices, "sensitivity" (the fraction of actually positive cases that is diagnosed as having the disease in question) and "specificity" (the fraction of actually negative cases that is diagnosed as free from the disease). Sensitivity and specificity represent, in effect, two kinds of accuracy: one for cases with a particular disease, and the other for cases without the disease. The terms "True Positive Fraction" (TPF) and "True Negative Fraction" (TNF) are sometimes used instead of "sensitivity" and "specificity." Then, "False Positive Fraction" (FPF) represents the fraction of actually negative cases diagnosed as positive with respect to a disease, and "False Negative Fraction" (FNF) represents the fraction of actually positive cases diagnosed as negative. (Note that, if all cases are diagnosed as either positive or negative, FPF = 1-TNF = 1-specificity, and FNF = 1-TPF = 1-sensitivity.) Because each of these indices represents a rate of accuracy or inaccuracy conditional on the presence or absence of disease, their values do not depend on the prevalence of the disease in question if the result of a diagnostic test (e.g., an image) is interpreted in a consistent way (1).

The sensitivity and specificity of a diagnostic procedure do not adequately measure the performance of that procedure, however, because the values of these indices are not unique; rather, they almost always depend upon the arbitrary selection of some "decision criterion" or "confidence threshold" (1). In the use of a simple quantitative diagnostic test yielding a single number as a result, such as blood serum assay, some "threshold of abnormality" must be selected, and the sensitivity and specificity of the test will vary inversely, depending upon the threshold adopted (1). Similarly, medical decisions based on diagnostic studies that are interpreted subjectively, such as medical images, require the arbitrary selection of some "confidence threshold": if the observer's confidence in a positive diagnosis exceeds some level, he will call the case positive; otherwise, he will call the case negative. (We assume here for simplicity that a diagnosis of either "positive" or "negative" must be made. In a real diagnostic strategy, some intermediate range of confidence in a positive diagnosis may be reported as "equivocal," with the suggestion that further tests be done. This strategy can be accommodated within the framework of the approach described here, but the corresponding methodology will not be discussed in this paper. The underlying concepts are the same.)

The appropriate selection of a threshold of abnormality for a quantitative test or a confidence threshold for an imaging study depends upon many factors, such as the prevalence of disease and (usually subjective) judgments of the relative advantages and disadvantages of the various possible kinds of correct and incorrect decisions (1). Thus, the selection of a decision threshold for interpreting the results of a diagnostic test depends upon the clinical context of a decision task and may vary from one decision maker to another because of different cost/benefit judgments. Because the sensitivity and specificity of a diagnostic test can vary inversely, depending upon the decision threshold selected, a single pair of values for these indices does not uniquely describe the detectability of disease provided by the test. Thus, despite their other advantages, sensitivity and specificity do not satisfy the need for a reproducible measure of diagnostic performance.

ROC ANALYSIS

This limitation of sensitivity and specificity as indices of diagnostic performance can be overcome by determining experimentally the various combinations of sensitivity and specificity that a diagnostic procedure can provide. The resulting functional relationship, or curve, is called a "Receiver Operating Characteristic" (ROC) curve and usually graphs TPF (i.e., sensitivity) as a function of FPF (i.e., 1-specificity). Because TNF = 1-FPF and FNF = 1-TPF, an ROC curve shows the trade-offs that are available among the relative frequencies of both kinds of correct and both kinds of incorrect decisions in a two-alternative decision task.

The ROC curve provides a description of diagnostic decision performance that satisfies all the criteria we listed earlier. By appropriate choice of the decision threshold, a decision maker or observer can operate at (or near) any desired compromise among the various decision frequencies that lie on the curve.

Figure 1 shows a typical ROC curve, together with three of the many possible operating points. The "strict" operating point might correspond to a situation in which the decision maker decides that a patient or image is "positive" if, and only if, he is quite certain that the disease in question is present. Then, FPF is small (i.e., few actually normal patients will be diagnosed as abnormal), but TPF is also rather small (i.e., an appreciable fraction of patients actually having the disease will be diagnosed as "negative"). The "relaxed" operating point, on the other hand, might correspond to a situation in which a patient (or image) is called "positive" whenever there is some arguable possibility that the disease is present. Then, TPF is high (almost all patients actually having the disease will be called "positive"), but FPF is also high (many actually negative patients will be diagnosed as "positive").

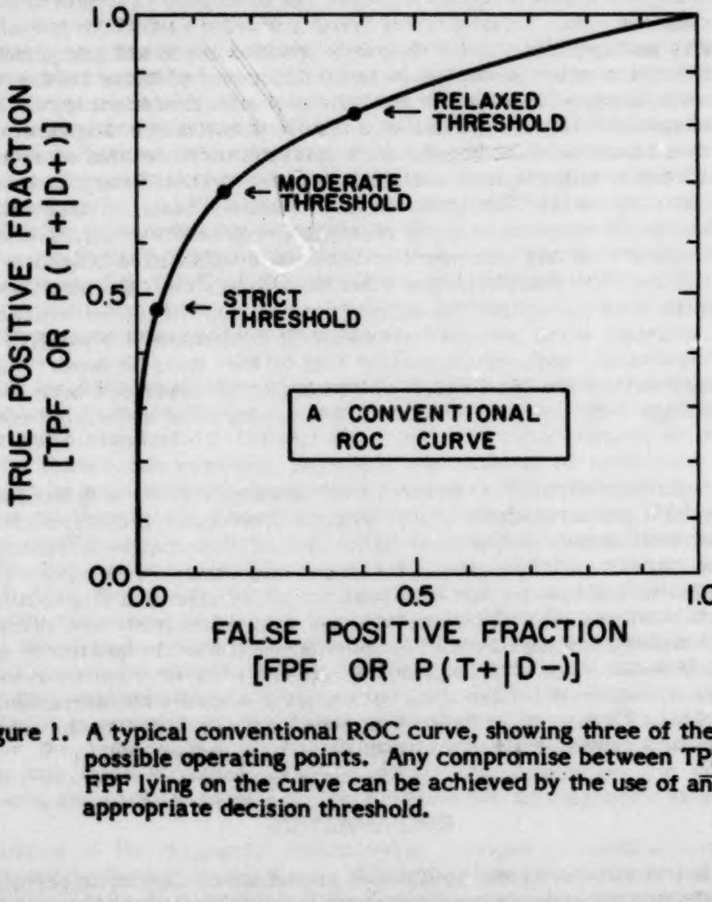

Figure 1. A typical conventional ROC curve, showing three of the many possible operating points. Any compromise between TPF and FPF lying on the curve can be achieved by the use of an appropriate decision threshold.

Essentially, a conventional ROC curve describes the compromises that can be made between TPF and FPF--and hence among the relative frequencies of true positive, false positive, true negative, and false negative decisions--as a decision threshold is varied for a given test. We have discussed elsewhere the factors that should be considered in selecting an appropriate compromise, or "operating point" on the curve (1). Since the ROC curve is a graph of TPF versus FPF, both of which are independent of disease prevalence when a fixed decision threshold is used, the ROC curve does not depend upon the prevalence of disease in a population to which the corresponding test may be applied. (The curve may depend on the spectrum of disease states included by the experiment designer as "actually positive," however. If early disease is harder to detect than advanced disease, for example, then the ROC curve will depend on the mixture of early and advanced, actually positive, cases studied. Thus, cases in the actually positive component of a study population must be chosen to represent the population at large to which the conclusions of the study will be applied. Similarly, the actually negative component should appropriately reflect the relative frequency of normal variants.) Thus, ROC analysis provides a description of disease detectability that is independent of both disease prevalence and decision threshold effects.

For diagnostic tests in which the test result must be judged subjectively, such as medical imaging, an ROC curve describes the decision performance of an observer/test combination. Clearly, disease detectability can be poor if the test provides little information, or if the observer is not skilled in interpreting the information provided, or both. Because it gives an empirical description of decision performance, ROC analysis of subjective diagnostic tests cannot reveal whether the technology or the individual human is performing badly. However, ROC analysis of the decision performance of several individuals using a single diagnostic test can indicate the extent to which usefulness of the test depends upon individual skill and/or experience (2). A more subtle issue related to performance of the decision maker, as opposed to the test, concerns his ability to hold fixed his decision threshold. Variations in use of the decision threshold from decision to decision cause decision performance to be degraded, with a consequent effect on the measured ROC curve (3). This effect of threshold inconsistency on the measured ROC curve is appropriate and desirable, because any aspect of decision-making behavior that degrades decision performance should be included in an empirical analysis of the observer/test combination.

EXPERIMENTAL DETERMINATION OF ROC CURVES FOR DIAGNOSTIC IMAGING PROCEDURES

Essentially, an ROC curve can be generated by varying the decision threshold that defines the "cut point" between results ascribed to (though not necessarily due to) actually positive and actually negative cases, and by observing the resulting changes in TPF and FPF.

Data from a diagnostic test that yields a single quantitative value for each case can easily be rescored as "positive" or "negative" by using various decision thresholds. A number of points on the corresponding ROC curve can be plotted in this way, and a smooth curve can be drawn through, or fitted statistically to, the points.

This approach is often impractical for diagnostic tests that must be interpreted subjectively, however, because human observers may not find it possible to associate a continuum of numerical values with their subjective impressions of certainty. The simplest way of expressing a diagnostic decision is in terms of "positive" or "negative," even though that decision may have been reached by comparison of a subjective impression with a decision threshold. These binary (two-valued: yes or no) decisions cannot be reanalyzed to determine what the decision maker would have said if he had used a different confidence threshold, however. Thus, an ROC curve can be generated from subjective "yes/no" response data only by requiring the decision maker to "re-read" the entire set of cases several times, using a different decision threshold each time. This repeated "yes/no" approach is clearly burdensome and usually impractical.

A practical technique for generating response data that can be used to plot an ROC curve in such a subjective judgment situation is the "rating method," developed in experimental psychology (4). Essentially, the method represents a compromise between accepting "yes/no" responses and requiring that the decision maker select a value from a continuous scale to represent his confidence that the case in question is positive. Instead, the observer/decision maker is required to select one of several "ratings" or categories of confidence to represent his judgment, based on the information provided by the diagnostic test (and perhaps on other supplementary information available to him). These categories can be given qualitative labels, such as: "definitely or almost definitely negative," "probably negative," "possibly positive," "probably positive," and "definitely or almost definitely positive." The use of five categories seems to represent a reasonable compromise between the needs of ROC analysis and the precision with which an observer can be expected to reproduce his ratings. Use of N categories will yield (N-1) nontrivial points on the ROC curve.

The rating scale data can be combined in various ways to calculate experimental estimates of operating points on the ROC curve (1,4-6). The uncertainty in each estimated operating point due to the combined effects of sampling statistics and observer variability can be described by binomial statistics and can be estimated by standard methods (1,4). A continuous ROC curve can then be fitted to the obtained operating points, either subjectively or by the use of a maximum likelihood curve-fitting computer program (7). We have published elsewhere an example in which rating scale data are used to determine an ROC curve (6).

TRUTH, CASES, AND COMMON SENSE

A fundamental aspect of almost any objective approach to the evaluation of diagnostic decision-making--whether in terms of accuracy, sensitivity, and specificity; or ROC analysis --is the need for a sufficient number of cases in which the actual state of health or disease has been determined. Diagnostic "truth" must be known in order to score the quality of each decision, and enough cases must be used to ensure acceptable statistical precision in the measured performance indices. Although these requirements are sometimes tedious to satisfy in clinical situations, ROC analysis is no more demanding in this regard than other objective methods of evaluation analysis. In short, the quality of diagnostic decisions cannot be determined if the correct answers are not known.

The problem of establishing "truth" is straightforward in evaluation studies that use artificial test samples or "phantom" images, but this problem can be exceedingly tedious and frustrating in studies employing actual clinical cases. The definition of "truth" is ultimately a philosophical issue, of course, and operational standards for diagnostic truth must be established for the purposes of evaluation analysis; these must take into account the goals of the evaluation study, potential sources of bias, and common sense. In short, standards of truth need not be "perfect" but must be considerably more reliable than the tests to be evaluated; judgments of truth should be independent from information provided by the tests to be evaluated (8); and one must balance thoughtful reflection on the potential errors and difficulties of such evaluation studies against the useful, even if limited, information that they can provide.

In the selection of cases to be included in an evaluation study, due consideration must be given to include an appropriate spectrum of disease characteristics in the sample case population, because the conclusions drawn from the study are applicable only to, and cannot be defined more specifically than, the sample population (8,9).

The various issues that should be considered in designing a study for the evaluation of diagnostic medical imaging procedures are discussed in a general protocol currently in the final stages of preparation by Bolt, Beranek and Newman, Inc., Cambridge, Massachusetts, under National Cancer Institute Contract N01-CB-64010 (Standard Protocol for Evaluation of Imaging Techniques in Cancer Diagnosis: John A. Swets, Principal Investigator).

GENERALIZED ROC ANALYSIS

The conventional ROC methods that we have described up to this point apply to situations in which actual states of health and disease are grouped into two categories and in which two decision alternatives are available to the decision maker. In this section we sketch how these methods can be generalized to apply to more complicated decision-making situations.

The most fundamental property of the ROC approach is that it describes the trade-offs that are available among the conditional frequencies of various types of correct and incorrect decisions. By viewing the approach in this broad way, we can see that a generalized ROC approach would account for the ways in which the frequencies of certain types of decisions must vary with the frequencies of other types of decisions as one or more decision thresholds are changed.

Consider first the situation in which the decision maker must not only call an actually positive case positive, but must also state where the case is positive in order to receive credit for a fully "true positive" decision. If localization of disease to within the proper image quadrant is required, then five actual states and decision alternatives are available: "no disease," "disease in upper-left quandrant," and so forth. We have shown theoretically and experimentally that decision performance in this more complex task can be predicted from knowledge of the conventional ROC curve measured for the two-alternative "detection-only" task, and that the resulting generalized ROC curve is a curved line in three-dimensional space, which can be plotted as two curves on a two-dimensional graph (10,11).

Another situation of interest is that for which more than one lesion, for example, may be actually present, and for which the observer must, in effect, count the lesions present. We have shown that if the possible lesions are similar, decision performance in this "multiple signal" task can again be predicted from knowledge of the conventional ROC curve (measured when zero lesions or one lesion may be present), and that the generalized ROC curve is a curved line in multi-dimensional space, which can be plotted as a set of two-dimensional graphs (12).

These two studies have shown that decision performance in some multi-alternative tasks employing medical images can be related uniquely and predictably to decision performance in a simple two-alternative task, which is measured by a conventional ROC curve. Thus, in these situations the conventional ROC curve provides a sufficient conceptual and experimental description of decision performance.

A common aspect of the tasks used in these two studies is that the decision maker can be assumed to base his selection of one of several decision alternatives on the repeated comparison of a single kind of judgment against a single decision threshold. In the "multiple-signal" detection task, for example, he is assumed to try to detect lesions in various parts of an image by repeating a similar judgment process and then "adding up" the number of lesions that he has "found."

An appropriate theoretical model for what we might call a "simultaneous detection and differential diagnosis" task is less clear (9). For example, suppose that the decision maker is confronted with a population of cases, each one of which may be actually "negative," "positive with disease A," or "positive with disease B." No fully general multi-alternative ROC approach is yet available to measure and describe decision performance in this task. An approach that may suffice at present is the measurement of three conventional ROC curves, either by grouping the actual states into two alternatives in the three possible ways, or by deleting cases with one actual state in each of three decision experiments.

Theoretical and experimental efforts to deal with this important situation within the context of ROC analysis are continuing.

IMPLICATIONS FOR MEDICAL DECISION-MAKING

In performing a diagnostic study, one pays a price (in terms of money, risk of complications, and/or radiation exposure) to gain information that should be of benefit in subsequent patient management. ROC analysis provides a means of measuring and describing diagnostic performance in terms of the combinations that can be achieved among the relative frequencies of true positive, false positive, true negative, and false negative decisions. Thus, through ROC analysis one can determine the information that a diagnostic test can provide. The term "information" here can be interpreted either in the loose sense of "detectability" or in the technical sense developed by Shannon (13,14).

With diagnostic performance specified by ROC analysis, several important questions remain, however. In a particular diagnostic task, which is the best of the possible combinations among the various decision frequencies--that is, what is the best operating point on the ROC curve? How can one judge whether the diagnostic information purchased by the use of a diagnostic test is (expected to be) worth the price paid? And how can a diagnostic test best be used within the context of a diagnostic strategy? These questions can be addressed, at least conceptually, by combining ROC analysis with the techniques of cost/benefit analysis and decision analysis. Discussions of this approach can be found elsewhere (1,15).

SUGGESTIONS FOR FURTHER READING

In addition to Reference 1, introductory discussions of ROC analysis for diagnostic evaluation have been published by Turner (16), by McNeil and colleagues (17,18) and by Swets (19), and these papers are recommended for the additional perspective they provide. Other introductory papers by Swets (20) and by Swets and Green (21) trace the development of ROC analysis in experimental psychology and indicate applications in other fields. We have published elsewhere a partially technical discussion of the ROC approach to diagnostic evaluation that includes examples of the various techniques (15).

An introductory book by Egan (22) clearly illustrates the mathematical relationships among various decision strategies, decision variable distributions, and the corresponding ROC curves. "Signal Detection Theory and Psychophysics" by Green and Swets (4) continues as the standard comprehensive reference work on ROC techniques. Finally, although it does not consider the implications of ROC analysis for optimizing diagnostic strategies, a classic book by Raiffa (23) provides an excellent introduction to the principles of decision analysis.

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CONSPICUITY EFFECTS IN CHEST RADIOGRAPHY

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WHY CONSPICUITY?

Radiologic systems and techniques are selected by people who on the basis of experience try to achieve maximal diagnostic usefulness at minimal cost, radiation exposure and patient discomfort. There is no uniform agreement about how this should be done and consequently there is great variation in clinical practice. Radiologic scientists have tried to narrow the range of disagreement by providing quantitative data about system performance. In general, they have concentrated on the fidelity of image recording and radiation dose. Parameters that express sharpness, contrast and noisiness at a given radiation flux have been developed and advocated as measures of what is commonly called "image quality." Attempts have been made to prescribe optimal radiologic systems on the basis of these parameters.

However, it has also been pointed out repeatedly that the objective of medical imaging is not to make accurate pictures, but accurate diagnoses. The optimal image is not necessarily the one of highest fidelity at lowest dose, but the one that best shows the diagnostic information. Conventional or computed tomograms may not have high image quality, and may have a high radiation cost, but may also contain diagnostic information not obtainable in conventional films.

Now, the nature of this diagnostic image information is not at all clear and at present it will have to suffice to define it as whatever is used from the image by an observer to make a diagnosis. This is not very satisfactory, but it does suggest an empirical method for determining which of two classes of images contains the most information: measure diagnostic accuracy. The technique with the highest accuracy has the highest diagnostic information content. Although studying observer performance might provide a method for choosing the best systems for use in clinical practice, the extensive clinical trials and the complicated methodology needed to get reliable data preclude its use for most clinical problems. An alternative is to develop methods of analysis that combine the exactness of physical measurements with the relevance of diagnostic information content as determined by observer performance. Kurt Rossmann was moving in this direction in his later work and this is the task we have been addressing for the past several years (12).

We have been attempting to develop a psychophysical relationship between quantities that can be measured in images and a quantity that can be measured in human observers: the probability of detection. This relationship, if sufficiently powerful, could be used to predict detection performance for specific abnormalities from physical measurements. Although there is a body of good work in this area, much of it is only marginally relevant because it deals with images that have very simple pictorial content (2,3,9): beads, wires or simple phantoms that only superficially resemble natural images.

The difficulty with natural images is that, because of their complexity, factors not accountable by measures of resolution, contrast sensitivity and noise may determine whether or not an abnormality is visible. In chest radiography, because of transparent overlapping, abnormalities can be camouflaged by normal anatomic structures. The ability to detect them may depend more on the geometry of projection than on image fidelity. Natural pictorial content, as against man-made phantoms, can either aid or impede the effective transmission of diagnostic information. An ambiguous abnormality may be perceived as part of a normal structure or may stand out prominently because of unique dissimilarity from normal surrounding structures. Visual search and visual pattern recognition may also be influenced more by natural pictorial content than by image fidelity.

The physical measurements must, therefore, account for--rather than neglect--the pictorial content, i.e., the meaning of the image.

To do this, we first consider that the image consists of features that are organized by the viewer into: a signal, the abnormality, and the background of normal structures or structured noise (as distinct from the more commonly discussed stochastic noise). The interaction between signal and structured noise determines how conspicuous of the abnormality is. Conspicuity is a quantitative relationship between the measurable properties of the abnormality that tend to make it visible and the measurable properties of the surround that tend to camouflage it. Essentially, it is a signal-to-noise ratio, where signal and noise are defined in terms of features which are perceptually relevant clusterings of image elements.

This approach is a departure from the usual attempts to characterize imaging systems in terms of parameters that only define the fidelity of contrast and detail reproduction which are intrinsic image, or imaging system, properties. It is being explored with the belief that systems cannot be compared or optimized without reference to the specific diagnostic tasks for which they will be used.

CONSPICUITY AND PULMONARY NODULES

Although the concept of conspicuity can be described easily in general terms, its validation requires a concrete demonstration. We have tried to apply it to pulmonary nodules; these abnormalities are relatively simple, yet clinically important. They can be simulated on chest films and used to produce test series for collecting data on observers' response to images with a natural appearance and the usual range of natural variability (8). We have studied both real and simulated nodules from the point of view of measuring conspicuity.

The image properties that we currently believe influence nodule detection most strongly are shown in Table 1. Some of these properties have been studied as independent variables.

Table 1.

Image Component	Property
Nodule	Size, shape
Nodule plus surround	Contrast, edge gradient
Local surround	Edge completeness, edge complexity
Distant surround	Global complexity

Size (5), contrast (11) and edge sharpness (4, 8) show direct relationships to detection. The effect of background complexity has been more difficult to quantify, although it is quite easy to demonstrate anecdotally that the visibility of a nodule varies with its position in the lung, and with projection (1, 11).

Nodules that are clearly seen on a PA projection can be invisible on the lateral and vice-versa. This is an obvious justification for taking two views at right angles. Less obvious is the fact that small changes in projection in a PA can also make the difference between visibility and invisibility. Most of this effect is due to the obliteration of the edge of the nodule by overlapping structures so that it merges with the surround. Figure 1 shows a nodule that by virtue of a few millimeter shift in projection merges with the hilar vessels. This is a gross illustration of a principle. More subtle changes occur clinically, but they do not reproduce well in photographs. The edge feature is also very important for separating a true nodule from a smudge on the film.

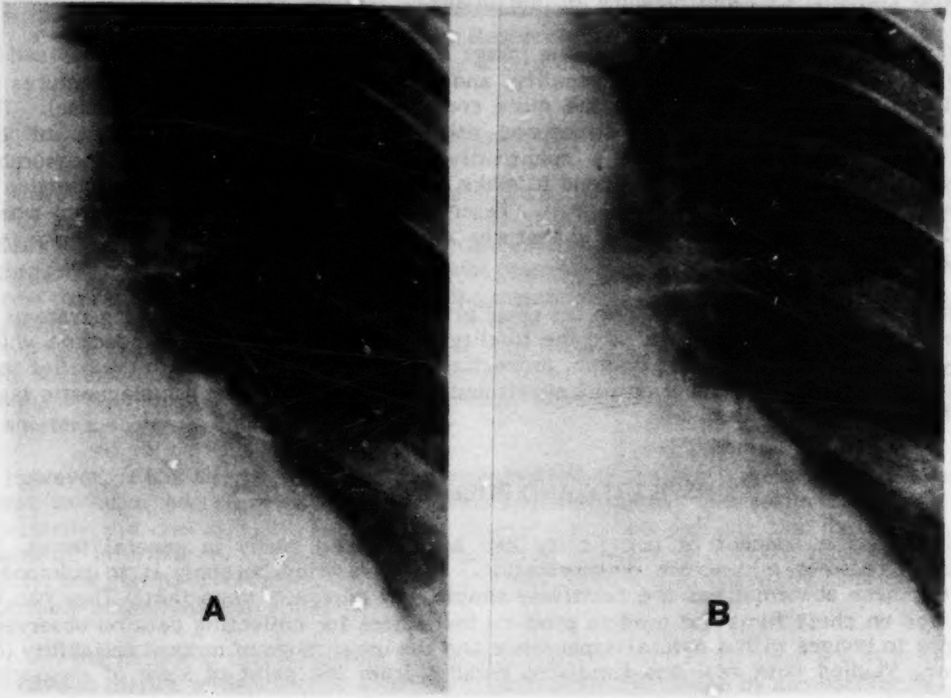

Figure 1. The effects of surround on detection. The clearly recognizable nodular density (A) becomes undetectable by a small shift in projection (B).

We have previously reported on a series of 62 cases with real nodules, where we had two films on each patient--one on which the nodule was missed by at least two film readers, and a later one on which it was detected (10). We were able to separate the two populations, missed and detected lesions, by using the following expression for conspicuity (K):

$$K = S \cdot \frac{\Delta D}{\nabla^2} \cdot EI \qquad (1)$$

where S is the nodule size, ΔD is the contrast, ∇^2 the surround complexity as measured by the Laplacian, and EI was an observer-determined edge index estimating both the sharpness and completeness of the edge.

At present, we are computing conspicuity as:

$$K = \frac{\Delta D}{b} \Big/ \frac{\nabla^2}{\sigma} \qquad (2)$$

where b is a measure of edge blurring, and σ is the sampling distance used to compute the Laplacian. The method of computation is shown diagrammatically in Figure 2.

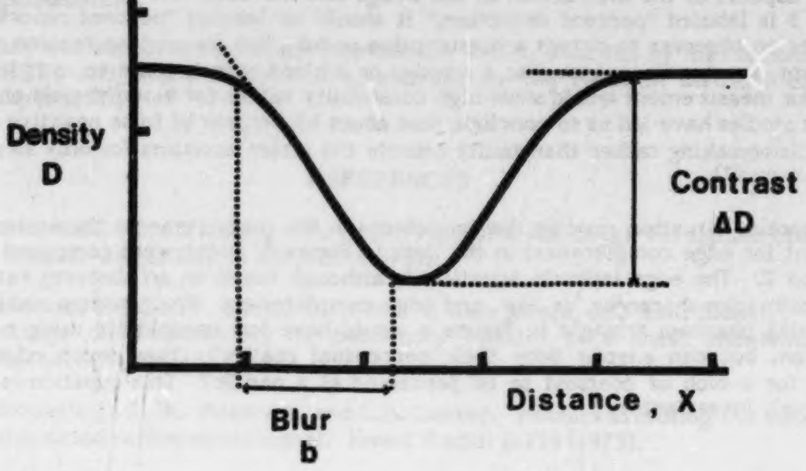

Figure 2. A method for quantitative description of edge blurredness. The line scan through a nodule center is used to define the blur b.

Figure 3 shows the relationship between the conspicuity of simulated nodules 1 cm in diameter with varying blur and the probability of detection as determined by 20 independent readings of the film. Most of the data are well behaved, but a few films consistently fall off the main curve. They are either inconspicuous nodules with high detection rates (open circles in Fig. 3) or conspicuous nodules with low detection rates (crosses in Fig. 3).

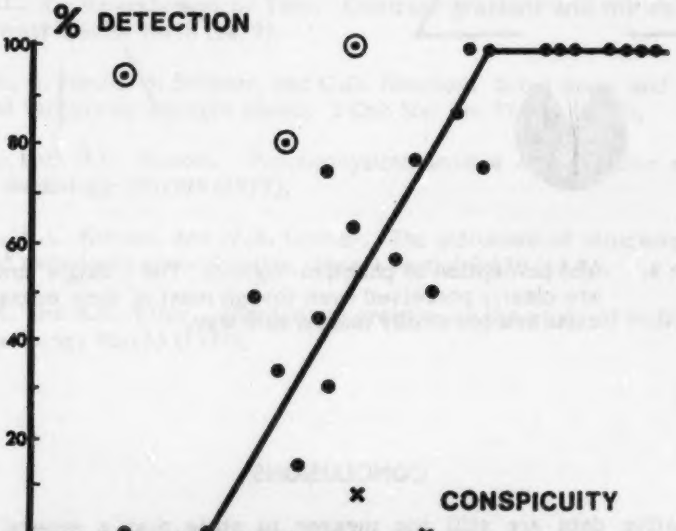

Figure 3. The probability of detecting a nodule as a function of its conspicuity. The data are based on 30 nodules viewed by 20 observer tests.

These films are particularly interesting because they draw attention to some often neglected aspects of the interaction of the image and the observer. Although the ordinate in Figure 3 is labeled "percent detection," it should be labeled "percent reported." It is possible for an observer to detect a presumptive nodule, but for various reasons not report it. It might, to observers, look like a smudge or a blood vessel, while to us it looks like a nodule. Our measurement would show high conspicuity values for blood vessels on end. Eye movement studies have led us to conclude that about 45 percent of false negative errors are due to decisionmaking rather than faulty search; the latter accounts for only 30 percent of false negatives (7).

The opposite situation may be due to defects in the measurements themselves. We do not account for edge completeness in the data in Figure 3, which were computed according to equation 2. The edge index in equation 1, although based on an observer rating scale, included both edge sharpness, or blur, and edge completeness. The phantom nodule and the more striking phantom triangle in Figure 4 would have low conspicuity using our present computation, but can anyone deny their perceptual reality? How much edge must be perceived for a blob of contrast to be perceived as a nodule? This question is currently being actively investigated.

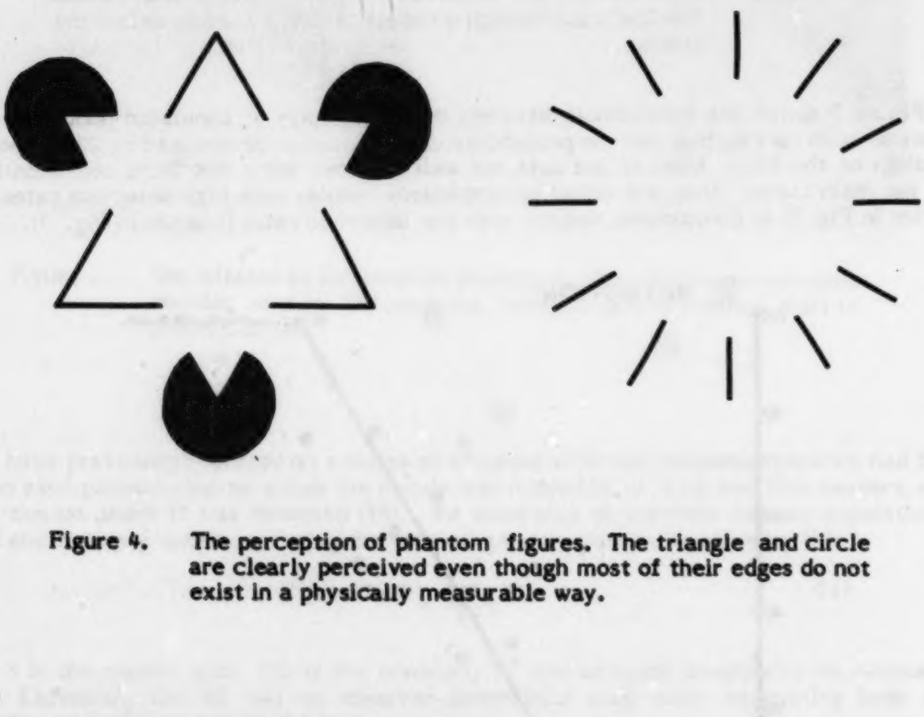

Figure 4. The perception of phantom figures. The triangle and circle are clearly perceived even though most of their edges do not exist in a physically measurable way.

CONCLUSIONS

The quantitative data are still too meager to state that a general psychophysical relationship has been developed that can predict observer detection performance from physical image properties. We believe that the preliminary data are encouraging and that continued investigation and debate on the validity of the concept of conspicuity may be very fruitful.

ACKNOWLEDGEMENT

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DISCUSSION

DR. MUNTZ: This comment is directed to Dr. Metz, but also to anybody else. As I listened to Dr. Metz's talk, I thought about climbing a mountain. There are generally several ways up. One may be a direct-aid route, and you hang the face for a while; the other may be a class-four route around the side, which is much easier and doesn't take very long. When Dr. Metz said we want to predict diagnostic performance, I wondered if we are really going about it the right way by working just on the image and measuring many image characteristics. Perhaps we should try to identify the dominant performance indices, which are also required for Dr. Metz's analysis and other types of analysis, and then optimize the image based on those indices being equal--rather than trying to look at the whole problem, which is like saying "How does this system perform for the whole range of all my performance indices?" I am sure that question is too big to answer in our professional lifetimes.

I feel we are trying to go the hard route. We also do not pay enough attention to modeling the examination or predicting what we see. For instance, Dr. Kundel's measurements are primarily off the image. They are associated only with the image. Trying to say what conspicuity is fundamentally, from a "how-it-got-made" point of view, would be more rewarding than simply making lots of different measurements over and over and over again for all possible variants of structural noise, or generating a new ROC curve every time you define a new task.

DR. TUDDENHAM: I would like to suggest to Dr. Kundel that his 60 percent decision errors may not all be decision errors. There is a real problem of attention. Attention is a significant part of perception. I think every clinical radiologist has had the experience of looking at a film and not remembering having consciously looked for disease. You drift off, and failure of attention may very well account for those lesions that are not perceived despite a person's fixing on the area for a prolonged period of time.

DR. KUNDEL: There is no way of knowing if you are right or wrong except that those patterns (of eye movements and fixations) that you saw were ten seconds' worth of viewing, so that you are paying an awful lot of attention, in terms of mind or body, to the area of the nodule.

DR. TUDDENHAM: By the same token, you may be very annoyed by the tracking device and be thinking about that when you should be concentrating on the film.

DR. KUNDEL: I thought you were going to allude to a much more important problem; that is, they may not be errors because what they do perceive may be too subtle for them to call. We have what we already considered a trade-off between normal variants and things that can look like pulmonary nodules and real nodules.

DR. TUDDENHAM: My point here is that there is more to perception than simply what is on the film. The attributes of the observer are important, and attention is one of these.

DR. GREGG: I would like to support both of Dr. Tuddenham's comments. There are some real data on the influence of attention. The work of Garland (L.H. Garland and A.L. Cochrane; JAMA 149:631, 1952 - Am J Roentgenol 82:25-30, 1959) and of Leo Rigler (L.G. Rigler; JAMA 142:773, 1950) showed that the true negative error rate in chest radiographs is around 25 percent. A second reading took it down to around 17 percent and a third reading to around 12 percent. All the ones that were missed were rather obvious, so it would seem to be a lack of attention. I am curious as to what this somewhat variable inattention really does to an ROC curve, which apparently does not take it into account.

DR. METZ: With regard to the question of whether or not ROC analysis applies to the evaluation of errors of search and attention, I want to emphasize that, in medical imaging, an ROC curve provides an empirical description of the diagnostic performance achieved by the combination of an imaging system and a human observer. Thus, the ROC curve includes

the effects of any and all human failings that limit diagnostic performance. However, if one wishes to study separately the effects of various possible physical and psychological causes of diagnostic error, then one can measure and compare ROC curves obtained under various experimental conditions.

Dr. Gregg suggested that false negative detection probabilities can be reduced by multiple readings or multiple images. Though I expect this is true, I feel obliged to point out that, in general, changes in false negative probability (or, equivalently, in the true positive probability) are meaningless unless one knows the corresponding change in false positive probability. I suspect that multiple readings or multiple images will increase true positive probability but will also increase false positive probability.

The effects of multiple readings or images on detection performance can--and should--be studied in terms of ROC analysis.

DETECTION PERFORMANCE AND THE NATURE OF THE RADIOLOGIST'S SEARCH TASK

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A chest radiograph presents a complex visual pattern; it is a two-dimensional distribution of the different x-ray attenuations produced by many superimposed structures. In viewing a chest radiograph, the radiologist often must be prepared to detect any of a large number of visual configurations that may reflect pathology. Such abnormalities include distortions in the size, shape, position or orientation of the silhouettes formed by normal structures (e.g., the heart, major vessels, bones and soft tissues), altered patterns of vascular markings, and any densities that are not part of the normal thoracic anatomy. The detection problem is complicated by the large number of possible abnormalities and the wide variety of normal patterns which can mimic subtle abnormalities.

Visual search is generally an integral part of the interpretation of a chest radiograph, but the search task varies with the clinical situation--i.e., the purpose of the chest film and the radiologist's prior knowledge about the patient. In some cases the radiologist may simply need to evaluate certain pre-specified features of the radiograph; in other cases, he may need to search the entire film for any abnormalities present. Between these two extremes, the clinical concerns may focus the radiologist's search within a particular region of the film or upon the signs of a specific type of pathology. How does the nature and complexity of this search requirement affect the radiologist's ability to detect abnormalities?

The answer to this question will depend upon how the detection performance is measured. If the measure is based upon the radiologist's ability to distinguish between normal and abnormal examinations, then the level of performance will appear to decrease as the search task expands and becomes more complex. An expanded search task increases either the area that must be searched or the set of abnormal conditions that must be considered. In either case, it increases the number of normal features considered on each radiograph, and thus increases the radiologist's opportunity to encounter a normal feature that mimics an actual abnormality. Even if there is no change in the radiologist's ability to distinguish between the individual normal and abnormal features on a radiograph, the statistical effects produced by an expanded search task will reduce his ability to differentiate abnormal radiographs from those that are normal. Statistical models can predict the magnitude of these effects from the number of additonal features that must be considered (2, 9)

In an experimental situation, it is possible to eliminate these purely statistical effects on detection performance by using a measurement procedure that keeps the observer's opportunity to make true and false detections constant for all search tasks. This procedure measures the radiologist's ability to distinguish the normal and abnormal features in a fixed "test set" that appears in all conditions. The experiments summarized here varied the nature and complexity of the radiologist's search task and investigated the changes in his underlying detection capability, as measured with respect to difficult sets of test features. Different search tasks varied the number of features on the radiograph that the radiologist had to consider in addition to those in the test set. The results show that detection performance does not remain constant, but depends upon the nature of the search task. However, the effects are the reverse of what would be produced by careless or faulty search procedures in the more complicated tasks.

METHODS

GENERAL DETECTION

One series of experiments (11,12) studied interpretation of a wide range of chest pathology on 27 radiographs selected from a larger set of 100 hospital films (5). The abnormal features in the test set were 47 subtle, but significant, conditions which would affect the patient's health or treatment. The normal test features were 82 artifacts, normal variants and superimposed normal features selected because they resembled significant abnormal conditions of various types. Eight radiologist observers read the films under three different conditions of search. In the free-search task they reported all abnormal findings on every radiograph. In the focused-search task they reported only the abnormal findings within certain designated areas of each radiograph, which included all the test features. In the non-search task, each of the 129 normal and abnormal test features was specified explicitly for the radiologist's consideration and evaluation. Confidence ratings were obtained in all three tasks.

NODULE DETECTION

Another series of experiments concentrated on the detection of pulmonary nodules (13). The abnormal test features were single simulated nodules on each of 48 different chest radiographs, 24 films that contained other types of abnormalities and 24 otherwise normal films. The normal test features were 222 specific locations in the lung fields of these 48 films, which contained superimposed normal structures that resembled pulmonary nodules.

Six observers read each radiograph twice, once with and once without the simulated nodule, in each of two search experiments. The initial, free-search task disguised any particular concern with nodules by mixing the 96 film presentations with an additional 48 abnormal radiographs, and by asking the observers to report all the abnormal findings on each film. They also indicated coordinates of each localized finding on the image, such as a nodule or a rib fracture, and gave a confidence rating for each abnormal finding. Several months later, the observers searched the films for nodules only, using the same reporting procedure but omitting all other pathology. In this nodule-search task, they tried to avoid omitting any actual nodules by reporting even low-confidence possibilities. The third experiment was a non-search task that specified each test location by horizontal and vertical coordinates on the image. The observer evaluated each specified location on the radiograph and rated his confidence that it contained a nodule.

PERFORMANCE COMPARISONS

The observer's performance was measured by the receiver operating characteristic (ROC) curve generated from his confidence judgments about the normal and abnormal test features in each experimental task (1,7). Such a curve indicates the proportions of both the abnormal and the normal test features reported above each of the various levels of confidence. The ROC curves for the same individual or group of radiologists, generated from the same fixed test set, were compared across conditions that varied the complexity of the search task, from non-search to free-search. Each ROC curve was fitted by a theoretical function whose estimated parameters characterized its shape and location (8, 11,12). A change in the ROC curve, from one interpretation situation to another, indicates a change in the observer's ability to distinguish between the normal and abnormal test features. This implies that the radiologist's detection processes must have been more effective in the condition that produced the higher ROC curve.

RESULTS

Figure 1 presents the group ROC curves generated from the pooled ratings of all eight observers, for the three experiments that studied the test set of general chest abnormalities. In order to make these observed ROC curves approximately linear, they are plotted in normal-deviate coordinates; the corresponding values of the actual true and false detection rates appear on the right ordinate and top abscissa, respectively. The solid lines represent the maximum likelihood fits to the group data points; the dotted lines represent the mean values of the linear parameters estimated for the fitted ROC curves of the eight individual observers. In general, the group performance will appear slightly inferior to the mean of individuals, because it has pooled together data generated by slightly different ROC curves and different decision criteria.

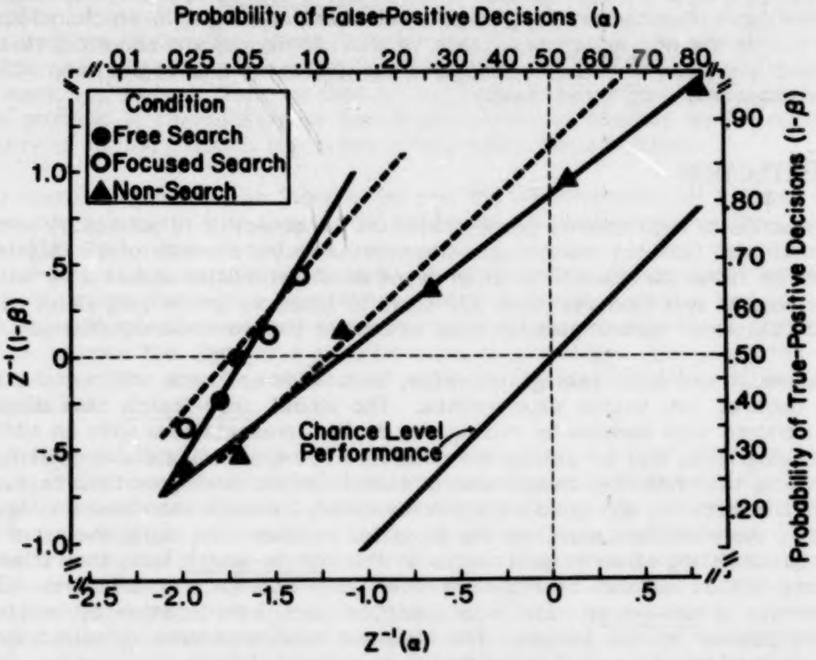

Fig. 1. ROC curves for the general test set from the pooled performance of eight observers in three search tasks, plotted in normal-deviate coordinates (12). The conditional probabilities of positive decisions for abnormal and normal test features (1-β and α) appear on the right ordinate and top abscissa; values of the standard normal deviates which yield lower tail areas equal to 1-β and α, $z^{-1}(1-\beta)$ and $z^{-1}(\alpha)$, appear on the left ordinate and lower abscissa. The two solid lines represent the maximum-likelihood linear fits to the group ROC curve in non-search task and the mean parameters of the fitted group ROC curves for free-search and focused-search tasks. Broken lines represent the mean values of the linear parameters obtained for the fitted ROC curves of individual observers.

The filled and open circles in Figure 1 show the group ROC curves from the free-search and the focused-search tasks. The only systematic differences between the two search conditions, in either the group or individual ROC curves, was a higher proportion of both true and false positive detection decisions at each level of confidence when the radiologist focused his search within specific areas of the radiograph.

In contrast to the similar detection efficiency in the two search conditions, the radiologists performed much worse when they did not need to search the radiograph for abnormalities. The triangular data points in Figure 1 show the group ROC curve from the non-search condition, which specified each of the individual test features for the radiologist's explicit consideration and evaluation. The observers needed to report much higher proportions of the normal test features in the non-search task to detect any given fraction of the abnormal test features. This effect was consistent for all eight individual observers.

Figure 2 presents the group ROC curves for the six observers in the nodule detection experiments, plotted in normal-deviate coordinates. Each data point indicates the proportions of the simulated nodules and the normal test locations on the 48 chest radiographs which were reported at or above each level of confidence in that task. Solid and dotted lines differentiate the fitted group ROC curves and the mean parameter values for the ROC curves fit to the six individual observers. The detection performance was best in the free-search task (filled circles), when the radiologists searched the entire radiograph for any and all abnormalities. Nodule detection was poorest in the non-search task (open triangles) when each test location was specified explicitly as a potential nodule. When the radiologists searched only for nodules (open circles), their detection of nodules was better than in the non-search task but inferior to their free-search performance. This ordering held for the ROC curves of all individual observers.

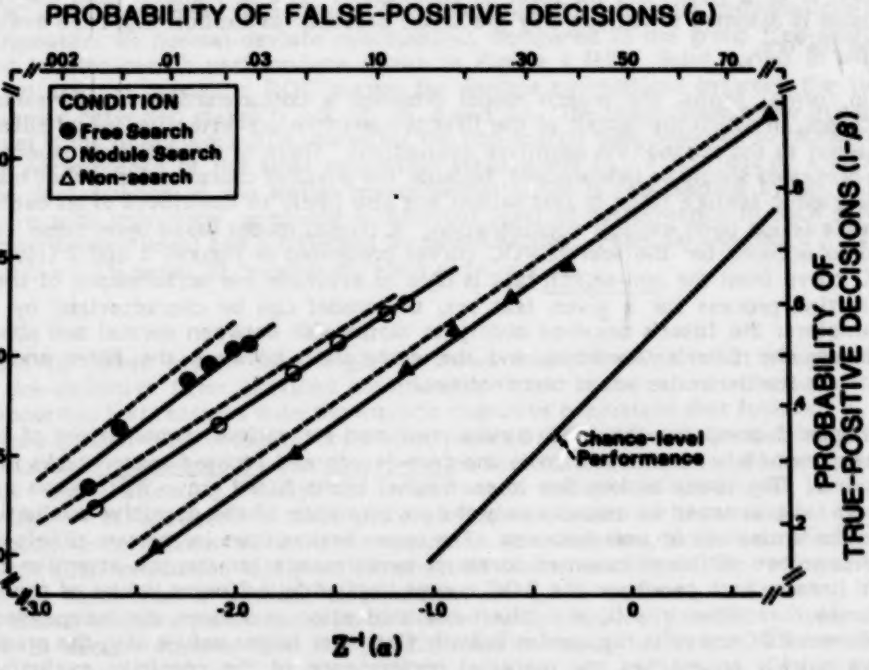

Fig. 2. ROC curves for the nodule test set from the pooled performance of six observers in three search tasks, plotted in normal-deviate coordinates (13). Solid lines represent the maximum-likelihood linear fits to the three separate group ROC curves; broken lines represent the mean values of the linear parameters obtained for the fitted ROC curves of individual observers.

DISCUSSION

The preceding experimental results seem highly counterintuitive. They indicate that the radiologist's detection processes actually function more effectively as the search task becomes more complicated, when he does not know precisely what abnormality to look for or where to look for it on the chest radiograph. The visual search of an experienced radiologist apparently does more than simply move his visual fixation and shift his attention from point to point on the chest radiograph. The search process also seems to involve perceptual mechanisms that substantially contribute to the detection of abnormal conditions, but which are unavailable for use when attention is focused upon a particular feature of the radiograph. Presumably, these are skilled "pre-attentive" mechanisms which an experienced radiologist has developed to draw his attention to features of a chest radiograph that are likely to represent thoracic abnormalities. For example, these visual mechanisms might be particularly sensitive to the presence of unusual textures, aberrant silhouettes and violations of bilateral symmetry.

A TWO-STAGE MODEL

The experimental results presented here can be explained by a model that treats radiographic detection as a two-stage process. The model assumes that when a radiologist searches a chest radiograph, skilled visual mechanisms automatically direct his attention to specific features of the image. These "pre-attentive" mechanisms function as a filter, selecting only the most salient features for explicit attention and evaluation. The radiologist's explicit evaluation of each selected feature then determines whether or not he reports it as a lesion and his level of confidence. In a non-search task, the radiologist's decision is assumed to reflect only the latter process, his explicit cognitive evaluation of a given feature.

In formal terms, the search model proposes a concatenation of two serial decision processes, in which the output of the first process (the pre-attentive visual filter) provides the input to the second (the cognitive evaluation). There is no reason to expect that these two processes should be independent, because the physical characteristics that make a given radiographic feature more or less salient are also likely to contribute to its evaluation as a possible lesion upon explicit consideration. A formal model based upon these ideas can be used to account for the search ROC curves presented in Figures 1 and 2 (10). When the ROC curve from the non-search task is used to estimate the performance of the cognitive evaluation process for a given test set, the model can be characterized by only three parameters: the filter's assumed ability to distinguish between normal and abnormal test features; the filter's threshold; and the correlation between the filter and evaluation processes for the entire set of test features.

Figure 3 compares the ROC curves predicted for various combinations of the model's three parameters to the data from the free-search and focused-search tasks presented in Figure 1. The lower broken line in each panel is the fitted group ROC curve for the non-search task, assumed to measure only the performance of the cognitive evaluation process over the entire set of test features. The upper broken lines in the two panels of Figure 3 represent two different assumed levels of performance for the pre-attentive filter. The solid lines in each panel are the ROC curves implied by different values of the correlation parameter r. When r = 0, the filter and evaluation processes are independent and the predicted ROC curve is high and relatively flat. For larger values of r, the predicted ROC curve quickly approaches the marginal performance of the cognitive evaluation, as this process is used to reject more and more of the originally selected features. The end point of each predicted curve depends upon the filter's threshold (C), which determines the detection performance when all selected features are reported. A change in the filter's threshold slides this end point up or down the line assumed to represent the filter's performance without much effect on its shape.

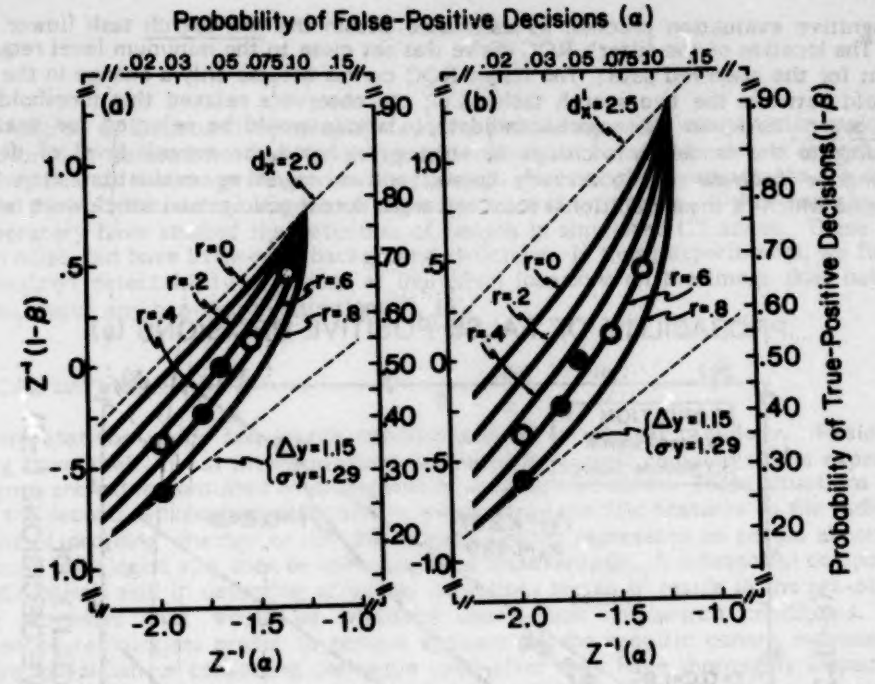

Fig. 3. ROC curves predicted by a two-stage formal model for specific combinations of parameters (in normal-deviate coordinates), compared to the group free-search and focused-search performance shown in Figure 1 (10). Solid curves in each panel are the predicted ROC curves for various correlations between the two assumed detection stages (r = 0, .2, .4, .6 and .8). The lower broken line in each panel is the fitted group ROC curve for the non-search task, assumed to represent the performance of the final evaluation stage alone (Δ_y = 1.15 and σ_y = 1.29). The performance of the initial pre-attentive filter is represented by an ROC curve with unit slope, assuming a different level of performance in each panel (upper broken lines) and a different threshold setting. Panel (a) assumes d'_x = 2.0 and C = 1.25; panel (b) assumes d'_x = 2.5 and C = 1.20.

To account for these search data, Figure 3 shows that the model must assume the radiologist's pre-attentive filter provided considerably better differentiation between the normal and abnormal test features than the explicit cognitive evaluation that followed it. In addition, the pre-attentive filter had to "screen out" a large percentage of the test features; Figure 3a assumes it eliminated 90 percent of the normal test features and 23 percent of the abnormal test features. Finally, the model must assume a substantial positive correlation (e.g., r = .6) between the filter and evaluation processes for these test features. Such a high positive correlation is not unreasonable in this situation, since the radiologist had to base his final evaluation solely on the physical characteristics of the radiographic image, with no historical or clinical information about the patient. Focusing the radiologist's search within a particular area of the radiograph apparently affected only the criteria used to decide whether (and how confidently) to report a feature selected for attention.

Figure 4 shows the model's ROC curves fitted visually to the group free-search and nodule-search performance from Figure 2. The upper broken line shows the ROC curve assumed to represent the pre-attentive filter's ability to select among the test locations. It was assumed to be the same in both search tasks and to have a slope equal to that of the cognitive evaluation process, as estimated within the non-search task (lower broken line). The location of the filter's ROC curve was set close to the minimum level required to account for the observed data. The fitted ROC curves assume only a change in the filter's threshold between the two search tasks—i.e., the observers relaxed this threshold in the nodule-search task, so that more candidate locations would be selected for evaluation. According to the model, this change in strategy reduced the overall level of detection performance because the observer's less-effective cognitive evaluation then had to determine which of these additional locations were actual nodules and which were not.

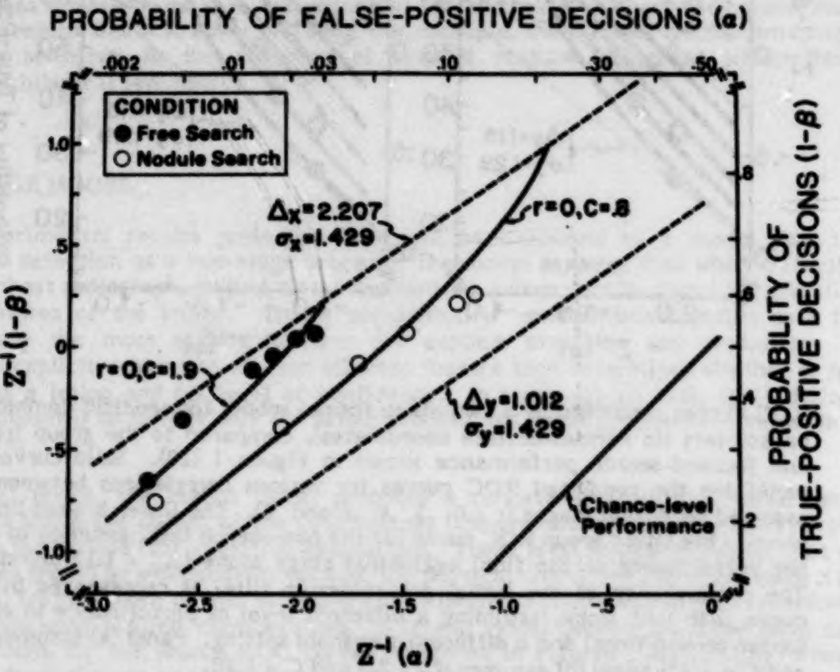

Fig. 4. Predicted ROC curves for the two-stage model visually fitted to the grou performance for the free-search and nodule-search tasks shown in Figure 2 (10 The assumed ROC curve for the initial filter stage (upper broken line) is assume to have the same slope as the curve for the final evaluation stage, as estimate from the fitted non-search performance (lower broken line). The group RO curves for the two search tasks were fitted by assuming a minimum level for th filter's ROC curve ($\Delta_x = 2.20$), the same correlation between stages ($r = 0$) an different threshold settings within the free-search ($C = 1.9$) and nodule-search ($= .8$) tasks.

The empirical ROC curves for all three tasks were nearly parallel, with approximate the same slope when plotted in normal-deviate coordinates. This forces the model assume that the filter and evaluation processes were virtually uncorrelated. A low or ze correlation implies that the selection of a test location for explicit attention depended up physical characteristics of the image that were quite different from those which determin its evaluation as a nodule or normal structure upon specific consideration.

The degree of structural complexity found in the radiographic examination may be the key factor that determines whether skilled observers will be able to detect lesions better when they search the image than when they do not. Normal chest radiographs are complex patterns that contain a high degree of organization and symmetry. Other experiments in our laboratory have studied the detection of lesions in simulated CT scans. These images contain noise, but have little or no background structure. In these experiments, we find that the measured detectability of lesions at individual locations on the image does not differ between search and non-search situations (4, 14).

CLINICAL IMPLICATIONS

Interpretations under non-search conditions arise in clinical radiology. Residents in training have their initial interpretations reviewed by a staff radiologist and experienced colleagues are often consulted on ambiguous or problematic cases. These situations tend to direct the second radiologist's attention to one or more specific features on the radiograph. In terms of deciding whether or not the alleged finding represents an actual abnormality, the second radiologist may then be operating at a disadvantage. A substantial component of the radiologist's skill in detecting abnormal conditions seems to reside in the pre-attentive search processes that would be bypassed under such non-search conditions. Many experienced radiologists prefer to remain unaware of the specific conern expressed by a referring physician or consulting colleague until after they have thoroughly inspected the patient's films and formed an independent judgment. Not only does this practice remove potential diagnostic biases, it should also yield improved decisions about the actual presence or absence of abnormality.

It is possible to reduce errors in radiographic interpretations by multiple-reading systems that combine the independent reports of two or more radiologists (3, 6, 15). The performance of such multiple-reading systems will depend upon the method used to resolve cases of disagreement between the original readers. In dual-reading systems, for example, these would be cases in which one reader of the pair reports a significant abnormal finding not reported by the other. We have distinguished two methods for using the opinion of a third radiologist to resolve dual-reader disagreements (6). In a "True Arbitration" method the third radiologist simply considers and decides each specific point in contention; in a "Pseudoarbitration" method the radiologist provides a third (search) interpretation of each film that led to a disagreement, resolving the specific dispute by a majority vote. The essential difference between these methods is whether the third radiologist interprets the film under non-search or search conditions. The empirical data presented here suggest that a Pseudoarbitration method should provide the most accurate resolution of disagreements, since it obtains an additional independent interpretation under search conditions. In a study of the dual-reader disagreements on 100 hospital chest examinations this method was able to retain 70 percent of the true-positive reports in disagreement, while eliminating 99 percent of the false-positives (6).

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STRUCTURED BACKGROUND EFFECTS ON OBSERVER PERFORMANCE

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University of New Mexico

Many investigators have reported that significant numbers of lung lesions are missed during clinical examinations (1-7). The results of most studies indicate that from 30 to 50 percent of the positive lesions may be missed during any single reading of a film series. Hessel et al. studied the changes in accuracy produced by multiple interpretation of chest radiographs in terms of effectiveness and cost (8). They found that multiple readings reduced overall errors. However, Yerushalmy found that two readers will disagree with each other in about 30 percent of the cases and a single reader will disagree with himself about 20 percent of the time (7). Thus, multiple interpretation of radiographs requires that the final answer be developed through some "arbitration procedure."

The quantum limited noise or random background so familiar in nuclear medicine images is not a problem in radiography. The most troublesome background in radiography occurs when local fluctuations of optical density (local features) obscure structures perceived by the observer. Note that the recognition of structures is determined by the observer and is not a characteristic of the variations in optical density in the image. Figure 1 shows a symbolic form of a triangle and Figure 2 illustrates how three additional unconnected features lead to the perception of a second triangle. Thus the observer relates the various features of the image to perceive real or imagined structures.

At the University of New Mexico we have been studying factors which influence the perception of simulated lung tumors. Our original tests were conducted using a series of positive radiographs created by a film duplication technique involving the combination of a mask containing a spherical lesion image together with a normal chest radiograph (9).

One of these was a pilot study designed to measure the effect of providing a second radiograph with slightly different surround. The lesion mask was shifted relative to the normal chest radiograph by 1 cm and a second duplicate test radiograph was made. Although the production of chest radiographs in this manner has some serious shortcomings because of the non-linear response of film, it did enable us to investigate the effect of changing the local background on observer accuracy, sensitivity and specificity (10). We found that, if the pairs of shifted films were divided into two individual series, the sensitivity and specificity results were the same for each series. We found that providing the film reader with the pair for side-by-side comparison improved the observer sensitivity from .42 to .57 (a relative improvement of 36 percent) and the specificity rose from .90 to .99 . There appears to be a significant improvement in observer performance when the immediate surround of the lesion is changed. A major benefit for the observer was the ability to make a side-by-side comparison.

We attempted to study the individual radiographs to see why some lesions were invisible on one of the pair but not the other and why some 10 to 15 percent of the positive films were missed by all observers. In some cases it was not clear why observers failed to detect the presence of the lesion. We were not able to successfully fit our data with the conspicuity factor as defined by Kundel (11, 12). It appeared that the presence and amount of structured background was an important parameter in determining the detectability of the lesion, but we have no clear idea of how to quantitate even the amount of background, much less the confusability of the background. The lung vascularity and rib shadows which normally appear on a chest radiograph were too difficult to quantitate in a reliable manner.

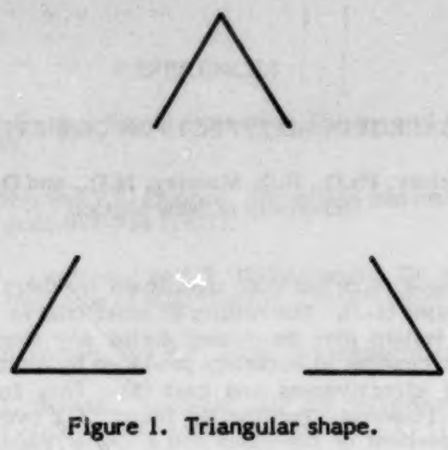

Figure 1. Triangular shape.

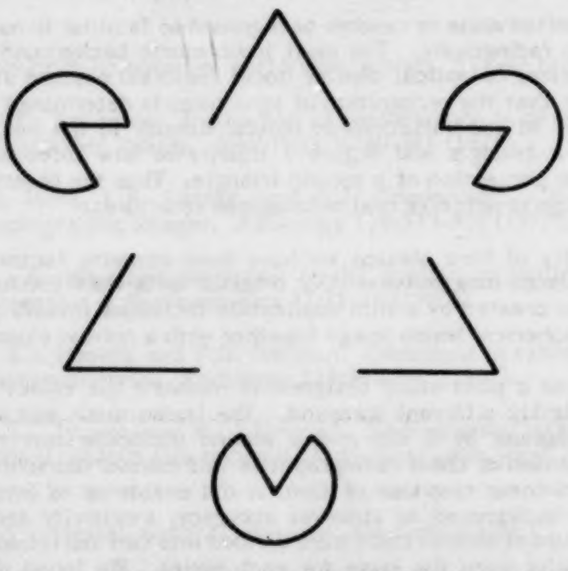

Figure 2. Triangular shape with added features leading to perception of second triangle.

We decided to step back from the clinical situation and look at simple targets with simple backgrounds and to begin our intitial studies by varying the amount of background.

The study was designed to investigate simple targets with simple backgrounds and measure the effect of changing the amount of background on observer performance as measured by sensitivity and specificity. Figure 3 presents a comparison of a task where the target is the same shape as the background, but a different size. The amount of background is the same in the two images. Figure 3a presents a regular arrangement and Figure 3b shows the same number of background objects and the same relative size of the target in a random distribution. There have been many studies done on visual search of clinical radiographs (13-16). It appears that radiologists are able to perceive major abnormalities during a very brief glance, and that many areas of a film are never searched by foveal vision. There have been several studies measuring the relation between observer performance and target and background size, as well as the confusion introduced by changing the size of the background structure (17, 18). Williams measured the number of eye fixations required to acquire a target (19). Bloomfield measured the time required to locate the target in a background of similar shapes (20). He found that the time to acquire and locate the target depends on the difficulty of the search situation, and that it is possible to develop an expression relating the probability of target detection in time "t" to the time required for a single search of the entire viewing field.

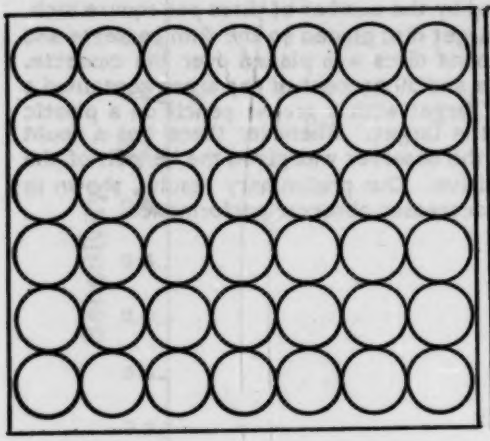

Figure 3a. Target in regular background.

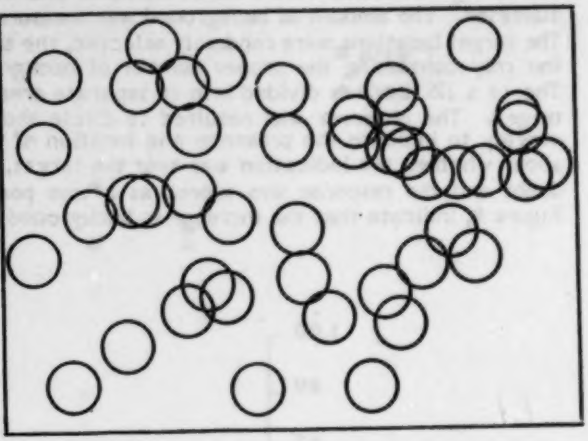

Figure 3b. Target in random background.

If P_{ss} is the probability of detecting the target during a single search, then $(1 - P_{ss})$ is the probability of not detecting the target after a single search, and $(1 - P_{ss})^N$ is the probability of not detecting the target after N searches. If the time for a single search is T_{ss}, then in time t there can be t/T_{ss} searches, and the probability of detecting the target in time t is given by:

$$P(t) = 1 - (1 - P_{ss})^{t/T_{ss}} \quad (1)$$

If the probability of detection with infinite search time is less than one, then equation 1 is reduced by the infinite search time probability.

This equation is consistent with our findings regarding radiologists' scan time and image search time in a clinical situation. Our experience indicates that most radiologists are willing to report a film as normal if an abnormality has not been found after three or four complete searches of the chest radiograph.

Engels has studied the effect of different shapes in a structured background field on the observer performance and measured a factor he defined as "conspicuity" for the different shapes (21). He determined the locus of points at which the target could just be detected when the eye was fixed on a central fixation point. The more conspicuous the target, the larger the angle through which the target could be displaced from the fixation point before detection ceased.

He defined the area of the locus as the "conspicuity." Not surprisingly, the targets which were most similar to the background were the most difficult to detect and were least conspicuous. We began a similar study in a radiology context to investigate how observer performance depends on:

(a) the amount of background,
(b) the relative size of the target and background, and
(c) the relative shape.

Our experiment consisted of presenting observers with radiographic images of wax discs. The background discs were 2 cm in diameter and the target discs varied from .7 to 1.7 cm in diameter. The amount of background was measured by the number of discs per square inch. The target locations were randomly selected, the target disc placed on the film cassette and the tray containing the proper number of background discs was placed over the cassette. The 14 x 17 film was divided into 16 separate areas and 50 percent of the areas contained a target. The observer was required to circle the target with a grease pencil on a plastic overlay to indicate the presence and location of the target. Whenever there was a doubt about whether the indication was over the target, the observer was given the benefit of the doubt and the response was scored as a true positive. Our preliminary results, shown in Figure 4, indicate that the increase in background decreases observer performance.

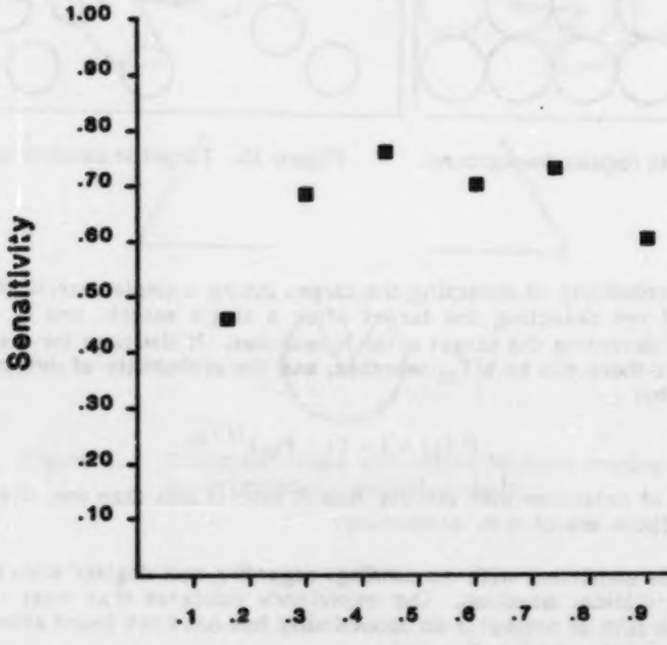

Figure 4. Observer sensitivity as a function of structured background density.

The dramatic decrease in observer sensitivity with a relatively small increase in background density suggests that there may be a threshold effect present. Figure 5 presents our preliminary results on a study of the effect of relative target-to-background size. The decrease at small ratios is surprising and will have to be verified by further studies. Future studies will include further investigations of a possible threshold effect in observer perception of targets in structured backgrounds, as well as of the effect of changes in relative target size and shape.

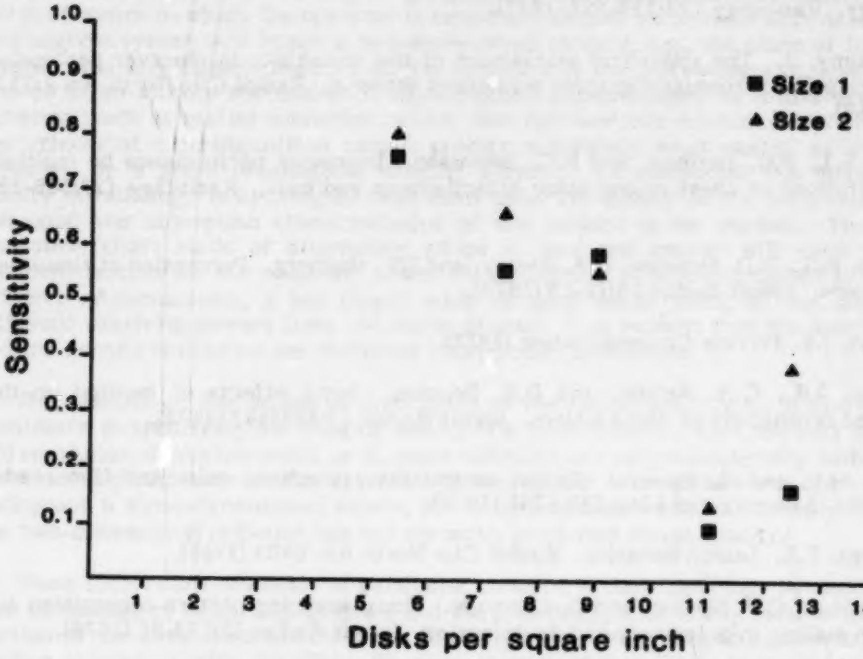

Figure 5. Observer sensitivity as a function of target-to-background size ratio.

ACKNOWLEDGEMENT

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RELATION BETWEEN 2D CRITERIA AND IMAGES OF 3D OBJECTS

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The classic criteria describing image quality in radiographs have been derived from the world of optics in which the operator is concerned almost exclusively with determining how well a given system will image a two-dimensional surface, e.g., the plane of focus of a lens system, or a very distant object, e.g., the surface of a star. A radiograph, however, may be likened to an infinite succession of superimposed planes imaged by a divergent beam; any criterion, such as spatial resolution, which does not take into account beam divergence and the effects of superimposition cannot predict accurately what quality of image will be produced by a given radiological imaging system. To compound this difficulty, image quality in radiology is also highly dependent upon the quality of the x-ray beam employed and upon the absorption characteristics of the subject to be studied. For example, a resolution chart made of alternating strips of lead and copper will yield two entirely different results for the resolving power of a system if it is radiographed at 20 kV and at 120 kV. Alternatively, a bar target made of bone would yield, at the same kV, quite different resolving powers from one made of lead. It is evident that the spatial resolution and the density resolution are therefore inextricably connected.

The question of the degree of divergence of the radiographic beam is also not currently considered in specifying the imaging ability of a given system. With the very thin (virtually 2D) resolution charts currently used, beam obliquity can vary considerably without changing the resolving capacity of the system and yet, if the two systems are then employed to radiograph a three-dimensional object, the images obtained will be markedly different, i.e., the two-dimensional criterion has not correctly predicted image quality.

These factors are obviously of particular concern in chest radiology because of the large size of the object being radiographed and the probability that in any view of the chest, that portion of the chest furthest from the film might be poorly imaged in comparison with the portion in contact with the film. We were concerned that there might in fact be so much loss of information from the distant portion of the chest that small lesions - including perivascular cuffing and loss of sharpness of vessel margins - could not be seen. This difference in imaging between the front and back of the chest could obviously not be described by a single spatial resolution figure - nor were we able to predict how much image degradation would be present. We hypothesized that using a minute focal spot would compensate for the poor geometry of the distant portion of the chest and we formulated a research protocol which permitted us to determine directly: (a) the effects of focal spot size on the relative quality (information content) of the images of the back and front of the lungs both on contact and on magnification radiography at various levels of magnification and (b) the relationship between the spatial resolution (using a conventional thin bar target) of a given imaging set-up (for example, 1.5X magnification) and the subjective estimates of image quality made by a group of observers.

An anthropomorphic chest phantom with blood-filled aerated lungs and a tissue-equivalent thoracic wall containing the bony thoracic cage was mounted on a 6-foot long optical bench. Several x-ray tubes with a wide range of different focal spot sizes were mounted in turn at the other end of the optical bench and radiographs of the phantom were made - on contact and with magnifications of 1.5, 2.0, 4.0 and 6.0 times. For each particular focal spot size and geometric arrangement the spatial resolution was measured at the most distant portion of the chest. A group of five observers of varying skill in interpreting radiographs was asked to evaluate the image quality of each film, scoring it from 1 to 10 (10 being the best quality). The observers were instructed that they were seeking a film which would be optimum for the detection of loss of sharpness of vessel margins and bronchial walls, i.e., the image quality was assessed for a specific task - an important point. The image quality scores for each focal spot size and geometry were then graphed against the spatial resolution derived from the bar pattern for that same imaging set-up.

The results showed that for a contact 6-foot AFD chest radiograph the single measurement of spatial resolution matched very badly with the observers estimate of image quality; i.e., there was considerable variation in spatial resolution according to the change in focal spot size, but virtually no detectable difference in image quality to the eye of the observer. We were able to section our phantom in the coronal plane and therefore to radiograph individual sections of the lung in space and compare the image of the most distal slice with that of the slice closest to the film. This part of the study showed that although there was quite marked degradation of the image of the distal slice as focal spot size increased, this degradation was entirely concealed by the superimposed image of the slice in contact with the film (which, of course, did not vary appreciably with focal spot size). Only when focal spot size reached 2.0 mm or more was definite image degradation detectable by the observers. We also found that, at a 6-foot AFD, focal spots smaller than 0.6 mm did not produce an appreciable improvement in image quality to the eye of the observer. As the degree of magnification was increased, it was interesting to note that a progressively closer match between the spatial resolution and the observers' subjective analyses developed until, at 4X magnification, there was an almost perfect correlation. Obviously, at this level of magnification the difference in resolution between the back and front of the chest is very much less than with a contact radiograph and the single resolution chart measurement is therefore much more predictive of the image quality that will be obtained.

We would suggest that this work is a further illustration of the value of using three-dimensional anthropomorphic phantoms in the study of radiologic imaging quality rather than relying on conventional physical parameters derived from objects which are virtually two-dimensional.

DISCUSSION

DR. SWENSSON: I hope that Dr. Milne does some observer detection studies to complement the very interesting observations that he has been getting.

DR. MILNE: So far I have been working on a more experimental approach, trying to achieve valid results that I could immediately transfer to chest radiographs, but I agree that the work is really not strong enough until the ROC approach is used on this data. You made a very important observation this morning that did not stand out in the large amount of data you presented and I wondered if everybody caught it. I am referring to your observation that, after a search had been directed, the subjects lost the ability to recognize objects elsewhere in the lung. This correlates closely with something that I am sure has happened to all of us. When you are driving rapidly along the road, suddenly something will strike you out of the corner of your eye, perhaps a restaurant name that you were told about 2 weeks ago. Similarly, when you are flipping through the pages of the telephone book you will see the name of a friend you have not heard from in 5 or 10 years. You can only have seen that name for a nanosecond as the pages flipped over, yet you picked it out of that page of a thousand words with instantaneous recognition. If you tried to go back and find the name, it would take several minutes of poring over the page before you could recognize it again.

I think that is analagous to the phenomenon you described this morning and may be a partial explanation for the apparent improvement in searching when the search was not directed; that is, when it was a general search, as opposed to the observer being told that there was something in a certain area. You have all had the same experience in front of the viewbox - you finish looking at a film and somebody says over your shoulder, "Yes, that is very good, now tell me the other two things that you missed." From then on, you have had it. You can search that film as much as you like and you will never see the other two things - even the fact that there is a four-quarter amputation and a hiatus hernia behind the heart! What is it that inhibits the continued efficiency of the search process? You make the eye movements; you know you are looking for something; but you cannot recognize it nearly as well as you could have 20 seconds before when you had no direction.

DR. SWENSSON: I would like to add something to that. I have never met a radiologist without a story about some film that he passed as normal but which contained a lesion that was embarrassingly obvious when subsequently pointed out by a colleague. As a result of these incidents, which I think are relatively rare, radiologists tend to believe that the overlooked finding is a frequent occurrence. When it happens to you, you remember it, but our data suggest that most features are "overlooked" for good reason. If these features were not overlooked, the rate of false positives would greatly increase.

DR. RAO: Dr. Milne mentioned that the two-dimensional projection of a three-dimensional object presents serious problems. Is not the variation among projections an equally important problem? Since anatomical structures within the body do not maintain definite geometrical relationships to each other, different projections can give rise to different contours and line patterns on films even if the technologist positions the patient in a standard manner. We heard from Dr. Ravin that small densities, such as lines and contours, are useful findings.

DR. MILNE: I would agree completely with Dr. Rao. I have been impressed recently by the difference between films taken of a patient in the intensive care unit positioned as close to vertical as possible, and films of the same patient taken perhaps 20 minutes later in the radiology department. The geometry is different and the technician has been trying to duplicate exactly the positioning, yet there is a major difference in the ability to see information on the two films. That is just an example of what you are saying: small projection differences, small differences in obliquity, and so on, can produce substantial changes in the radiograph. I think Dr. Ravin showed that very nicely with one of the nodules he made virtually disappear by a slight difference in off-centering.

Yes, I think that is a major problem and it is the principal cause for the lack of success in using subtraction views of the chest, because it is so difficult to get all the structures in exactly the same position on two successive radiographs. I am not sure there is anything very much we can do about it. I was interested to see that there is nothing in this program about stereoradiography and nothing about the effects of parallax. I do not know if anyone will be talking about it but, obviously, the positional problem would become much less important if every radiograph were a good stereo representation, because then you would not be concerned by the slight differences in projection. I would be very interested to know if anybody has tried to correlate information abstracted from stereo views of the chest with that from straight PA views in order to assess whether stereo adds any information.

DR. KUNDEL: About 15 years ago, Berkson and his colleagues at the Mayo Clinic did a study of stereo versus non-stereo chests for detecting lung tumors (Am Rev Resp Dis 81:660, May 1960). They did a nice study and their operating points can be plotted on an ROC curve. The stereo turned out to be better than the non-stereo. The problem is that they did not use the crucial control in that study. They compared a single film to a stereo pair but they did not present two films as a non-stereo pair. So it is very difficult to know whether the stereo, or just having a second film at a slightly different projection, was responsible for the improvement.

Human observers are very adaptable. Whenever you study perception, you discover that in one way or another you are studying learning; the "observers" learn to expect what you want of them and to do it. Our department routinely stereos everything except chests and we have learned to determine whether a structure is anterior or posterior to something else by just looking at the two films monocularly, without even resorting to stereo, because we have learned the relative effect of "stereo shift" between the two projections.

DR. KELSEY: We are in the middle of a study on that right now. We have a series of clinical films on which the presence of nodules is "proven" either by CT scans or linear tomography and we are doing exactly what you suggest. We present each film individually, then side by side, and then in stereo. We hope to have the results by the November 1979 RSNA meeting in Atlanta.

DR. KUNDEL: I might make a comment about models. There has been some talk about the deficiencies of the various models used to study this process. I think that each model has its deficiencies and its advantages. The important thing is to choose an appropriate model for the problem under study. The purpose of the experimental model is not to perfectly imitate reality but to catch the essence of some problem so you can abstract a general truth. I think it would be very difficult to use an anthropomorphic phantom in the kinds of studies we are doing because we need hundreds of different films and we would have to have a phantom that has not only dozens of different lung configurations but also dozens of different rib configurations. We would need literally dozens of phantoms and I do not know whether you appreciate the work that has gone into producing a phantom that looks real. This is not something that is simple to do and I do not think it is ready for mass production yet. We would need many phantoms with many different configurations. There are many places for each one just as there are places for bar and test patterns. These are not rival systems. We are just beginning to be able to look at the problem and acquire quantitative data about the perception of these things. I think that is the most important thing.

DR. MILNE: I do not disagree. I was not in any way suggesting that the three-dimensional phantom is the only type of model that can be used. Far from it! What I am suggesting is that one has to be aware of the deficiencies of each of these individual systems and I would agree that the solid phantom also has a deficiency. As has just been pointed out, one of the real difficulties is the time it takes to create one. It does not take as long, however, to create the various lungs to fit inside that phantom. So, it is quite possible to have some 20 to 30 different "pathology" configurations. In fact, just within the day, it is possible to change the nodule or interstitial patterns. I do not want to preempt the actual phantom part of the program by saying anything more about this because it will be addressed in considerable detail by Drs. Pearce and Gillan.

DR. GREGG: Dr. Milne, your data on focal spots show an obvious difference in resolution, including spurious resolution, between the X- and Y- directions. Have you noticed any effect of this clinically? The second question I wish to ask concerns the use of modulation transfer functions (MTF) that have a rise at the highest frequencies, namely Xeroradiography. Has anybody found this to be an advantage in chest radiographs and, if so, under what circumstances?

DR. MILNE: Let me take the second question first. Yes, Xeroradiography has a real advantage in visualizing the peripheral vessels of the lung. As a chest radiologist, I would use xeroradiography almost exclusively if it were not for the high radiation dose. It has the advantage of smoothing out the contrast gradient so you see details in the chest wall, ribs, mediastinum and lungs. It is really quite a superb technique; if the radiation dose could be reduced, I think I would be using xerography very frequently.

The first question was on two-dimensional variation in MTF. No, we have not been able to show it in the lung. In fact, we have to work rather hard to demonstrate it at all on clinical radiographs, though we finally did succeed in the neck of the femur. The neck of the femur is a natural bar pattern whose trabeculae are oriented parallel to each other and we were able to produce spurious resolution and "doubling" of the trabeculae in the neck of the femur simply by turning the x-ray tube until it was perpendicular and then diagonal to the trabeculae.

DR. SWENSSON: I am not sure I know exactly what you meant when you said a "rise" in the MTF. Is what you are talking about merely a roll-off of the low-frequency response?

DR. GREGG: It is the complete absence of low-frequency response.

DR. SWENSSON: Then that is analagous to what happens in the visual system: there is a low-frequency attenuation (or attenuation of low-frequency information). Whether this form of filtering is an advantage depends on what you are looking for. Filtering out the very low frequencies enhances fine details and edges but reduces information about gradual changes in intensity. In chest films, you have to judge for yourself whether this kind of information has any clinical relevance.

DR. MUNTZ: I would like to ask a general question. The important thing to any of us in optimizing anything is to have an idea of what variables govern the work we are doing. In other words, what are the performance indices? Presumably, there are a number of these, depending on circumstances. There is really not too much effort going into developing scientific ideas of what these indices might be. In order to optimize an entire system, understanding what parameters enter into the performance is of fundamental importantance.

DR. KUNDEL: Could you give me an example of a performance index?

DR. MUNTZ: Well, a simple one is some sort of signal-to-noise ratio.

DR. SWENSSON: I do not want to preempt Dr. Kundel's answer, but first I think you had better specify what performance you are talking about. What task is the observer supposed to be doing?

DR. MUNTZ: How do you generate an ROC curve? You assume a performance variable.

DR. SWENSSON: That is the point. Your measurements relate specifically to the particular task the observer performs which determines what information he needs from the images. Since the measurements you get depend on the kind of task you pose, the physical predictors of these performance measures are also going to be specific to that particular type of task.

DR. MUNTZ: To connect the rest of the examination--the resolution of the detector, the size of the focal spot or the properties of the subject--to the image and then to optimize that image for dose, cost, or whatever, you have to know from a systems analysis point of view the index you are optimizing. The inputs to your perception studies are also performance indices and you do not explicitly state them. That is the problem. Is it possible to look in more detail at the connection between your ROC curve and what is generating that curve in terms of a performance index or the diagnostic variables involved?

DR. KELSEY: That, of course, is what we started to do by stepping back and asking how we can quantitate the structured background. How does this depend on shape? One of the pluses is that you can quantitate these things, the minus is that you are a long way from the clinical situation. We recognize that.

DR. MUNTZ: I cannot do much with numbers of "discs," so I guess what I am looking for is some more fundamental measure - a conspicuity type of thing where you have a Laplacian of the background or at some frequency a contribution to the power spectrum that is extraneous to the signal.

DR. SWENSSON: I would like to respond to that. I think what you are asking for is what Dr. Kundel is trying to do with "conspicuity"; that is, to develop an appropriate measure--analogous to signal-to-noise ratio—that describes the physical characteristics of the image and predicts detection performance. In our laboratory, we are looking at the actual physical signal-to-noise ratio on simulated CT scans where, in fact, it is a good predictor of performance. However, that is a situation in which detection performance is fundamentally noise-limited. In the chest film the limitation is not noise at all, it is superimposition. I do not know how to go about quantifying that.

DR. MUNTZ: Of course, the big question is what you do to describe your camouflage. There certainly must be a lot of classified literature on this subject.

DR. METZ: I personally feel there is no shortage of indices, as you call them, floating around the world. But, a real shortage exists in studies that examine the extent to which prediction of these indices correlates with measured observer performance. I once counted all the different definitions of "signal-to-noise ratio" I could find and there were at least a half-dozen; the predictions of these various indices often disagree. I do feel fairly strongly that, as Dr. Swensson suggested, the indices appropriate in various situations are probably completely different. One index is probably appropriate in a noise-limited situation, another in a contrast-limited situation, and so on. There is not going to be any one magic formula, at least not in the foreseeable future, which will predict observer performance. I think we lack a bridge between the physical properties of imaging systems and the way the various components combine to predict an image, on one hand, and empirically measured observer performance, on the other hand. Work has been done in both these areas, but we have to start building the bridge between them.

DR. KUNDEL: I would like to reiterate what Dr. Metz just said. I think our theoretical understanding of the problem and our ability to measure things are so primitive that we cannot answer the kind of questions you ask. For example, when we try to measure contrast, such as the contrast of a pulmonary nodule, just start thinking about how we are going to measure this and what we are going to measure. There are 10 or 12 different definitions of contrast and each one produces a completely different curve when you plot it. The measurements themselves depend upon underlying formalisms, many of which may be wrong. For practical purposes, the best way to determine the optimal image is to get a consensus of people who actually use these images. Radologists are tremendous optimizers. What we are trying to do is find out how they do it. The best we are going to be able to do is tell how they actually come up with their good decisions. I think they do an excellent job of this kind of optimization without knowing the parameters involved.

One problem in psychology is that everybody has a mind and a brain and when you study it, you show a beautiful, psychophysical relationship which was intuitively obvious all along. If the relationship is not intuitively obvious, there is probably something wrong with it. Our state of understanding of the perceptual and diagnostic process is much too primitive to be able to reduce it to a signal-to-noise ratio. That would be an oversimplification of a hopelessly complex problem.

DR. MILNE: I think you are being a bit unfair, Dr. Kundel. Using "primitive" technique, we have still shown some things that, though simple, are useful in design of apparatus. I might give you some very simple examples.

If you want to image an abscess of the liver, there is no use taking a radiograph even with a magnificently minute focal spot and high-resolution screen/film combination, because you still will not distinguish the density differences between liver and abscess. But that abscess can be imaged if you use a system that provides good density discrimination, even if its spatial resolution is terrible. In the same way, if a small lesion in the lung has a very large amount of overlying structure and you tomograph it, the tomographic system, even with poor spatial resolution, removes the surrounding and overlying structured noise from this lesion, resulting in a good image of the lesion. Having that, you can build better spatial resolution into the system. Let me give you a further example. If you try to radiograph the ossicles of the middle ear with an intact skull, you will not be able to see the joints between them. If you then tomograph them, you get a much better picture of the joints between them, even with a system having poorer spatial resolution. But, now, tomograph them using a system with superb spatial resolution. You not only remove noise from in front and back of the ossicles but simultaneously get superb delineation of the joints between them. I could go on providing such examples, where there are very specific parameters we can measure and thus suggest how to optimize the image for certain diagnostic information. It is fair to say that sometimes the parameters we require for a good image are not known, but many times, as I indicated, they are. They just have not been used properly.

DR. GREGG: In my opinion, radiologists have done very well with limited equipment. What they have done has depended entirely on the available equipment. We have to look to the future and try to improve both equipment and techniques. Predictably, Dr. Milne, you will change the type of equipment you are using to look at different subjects, simply because we have noise-limited systems. You will always get more information with sharpest focal point, highest dosage, etc. So we must face the problem of dose limitation appropriate to the task at hand. The real question is whether we should design a whole series of expensive, special instruments, each adapted to a particular task, or whether we should have general-purpose equipment that may disclose most things wrong with the individual patient, excluding such complicating considerations as contrast media. I wonder if the panel would address itself to this problem in chest radiography: Would you design one system, utilizing low spatial frequencies, to look for lung tumors and another, utilizing high frequencies, for bone problems or vessel edges—with optimum dose for each task--or would you rather do it all on one machine?

DR. KUNDEL: Are you asking me as a radiologist or as a taxpayer? The difficulty is that there are costs incurred in two areas. There are radiation costs incurred in using no screens to get beautiful, high-resolution images at a very large radiation dose. Highly specialized equipment also costs in dollar terms, with the further problem of making it available throughout the community; then there are associated political problems.

We must know why the film is ordered and what are the patient's clinical problems. What clinical questions are we to answer? On the basis of the clinical questions, it may well be possible to say that one patient should have technique "A" to get the highest yield and that another should have technique "B" to get the highest yield. If we are looking for metastatic disease, it pays to get film that is going to optimally view the bones. We have to look at the clinical payoff and pay much more attention to the clinical aspect of radiography. We do not just do a chest film; we examine a patient for a particular purpose. The answer to your question is that I would like to have the flexibility to apply the most appropriate technique to each patient, which means I do not want a single, general-purpose machine, or a machine with one button on it that says "exposure." Rather, I would like to be able to vary technical factors and vary properties of the system to suit the task. I can, in an ad-hoc way, describe what kind of a system would be best, but we need to know a lot more about the payoff using various systems, given the clinical factors and the information we have about the patient.

DR. RAVIN: I guess we do have a bit of a philosophic difference. As I tried to indicate this morning, you are probably not going to be able to see everything on one film. As we practice radiology today, it is very difficult to do what Dr. Kundel suggests. If you look at case histories, they do not often tell you whether you should optimize for the mediastinum or the lungs. However, it ought to be possible to develop a system in which one can visualize, on a single film, both the mediastinum and pulmonary parenchyma. The major problem in developing such a system thus far has been the difficulty, because of the H & D curves of available films, of increasing the density in the mediastinum enough to allow one to optimally see the structures present, and at the same time not blackening the lung fields too much. I do not think the problem is insurmountable. I also do not think we have to know before we start whether we want to look at the mediastinum or at the lung because, in most cases, we will not know the answer. In terms of screening, we ought to be able to develop a system that would allow us to do both. I agree that we may not be able to optimally look at the bones on the same film, but that seems to me to be a smaller tradeoff. I would stress that there is really no technical reason, given current equipment and receptor systems, why we should not at least be able to see both mediastinum and lung on an "optimally" exposed film.

HIGH KILOVOLTAGE CHEST RADIOGRAPHY: AN HISTORICAL PERSPECTIVE

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Efforts to optimize radiography in general, and chest radiography in particular, date back at least to the turn of the century. Some proposals born of such efforts have been enthusiastically received and incorporated into the standard practice of radiology. Others, though seemingly rational in concept, have failed to win general acceptance. It would appear, therefore, that a meeting such as this might learn something from the experience of the past with respect to which approaches to improved chest radiography are most likely to prove fruitful. Others in this section of the program will undoubtedly present the rationale of high kilovoltage radiography in great detail. It is, therefore, my purpose to review the experience of the past with respect to a rather extensive effort to improve chest radiography by employing high kilovoltage radiation, launched over a generation ago, in which I was privileged to participate.

Let me take you back to 1950, when Roy Greening was working at the University of Pennsylvania and seeking to correlate ante mortem radiographic findings with autopsy findings by exposing radiographs of intact cadavera and their excised lungs during post-mortem examinations. These radiographs were compared with radiographs exposed shortly before death. A striking finding of this study was that, in several cases, large tumor masses were detected in post mortem radiographs that had not been identified in radiographs exposed shortly before the death of the patient (1). This observation prompted very serious questions: "How could such large lesions be 'concealed' in conventional radiographs, and what might be done to obviate this problem in the future?" Recognition of the rapidly increasing incidence of lung cancer lent urgency to these questions. Accordingly, a group was formed within the Department of Radiology at the University of Pennsylvania, under the preceptorship of Dr. Eugene Pendergrass, to investigate this problem.

It was decided initially to seek to define what characteristics of the radiographic shadow determine its perceptibility under conventional viewing conditions. Plexiglas test objects of precisely milled contours and dimensions were produced and so recorded that by presenting a population of these radiographic images to observers under controlled conditions, it was possible to study the effects of size, sharpness, and contrast on the likelihood of an observer's perceiving an isolated x-ray shadow in a uniform surround. Such studies showed that, at the visual threshold, the diameter of a shadow (between 2 and 40 mm) had no significant effect on the likelihood of its perception. Moreover, neither its contrast nor unsharpness alone appeared to determine the likelihood of a shadow's perception. However, contrast divided by unsharpness, i.e., the density gradient across its border, did bear a direct relationship to the likelihood of one's perceiving a shadow. A plot of these data, showing that as the density gradient across the border of the shadow increased the likelihood of its perception increased, is noted in Chamberlain's Carman Lecture of 1952 (2).

It appeared that, given a fixed reading distance, the perception of a shadow depended ultimately on the rate of change of density (and hence of luminance) as a function of distance across its border (3). This was a gratifyingly fundamental observation, but it was difficult to relate to the radiologist's failure to detect significant lesions in clinical radiographs. The problem of identifying an isolated target in a homogeneous field bore little relationship to the problem of identifying a clinically significant shadow in the tremendously complex visual display of the chest radiograph. Ultimately, on largely empirical grounds, it was concluded that lesions might be obscured in conventional radiographs by at least three different mechanisms. First, they might be obscured because of the relative underexposure of the mediastinal shadow, which in most radiographs was clearly on the "toe" of the film characteristic curve. As a result, regardless of the subject contrast between mediastinal structures, little mediastinal detail could be recorded. Second, it appeared likely that small peripheral pulmonary lesions might be completely obscured by the radiopacity of overlying ribs and, third, it was argued that superimposed shadows, even though they did not obscure a small peripheral lesion, might limit its perception by breaking up its coherent, continuous contour. Later studies confirmed the importance of the effect of such "background" shadows and showed that the perception of a target was strongly influenced by the contrast of the background pattern (4).

Two approaches to the solution of these problems were considered. First, the exposure in a single cassette, but between screens of different speeds, of two different films was proposed to provide one radiograph of proper density for the appreciation of lung field detail and a denser radiograph for the appreciation of mediastinal detail. Second, it was proposed to employ very high energy radiation in an attempt to exploit the fact that its absorption (largely by Compton attenuation) would be less dependent on the effective atomic number of the absorber than would the absorption of conventional diagnostic x-radiation. Calcified structures should, therefore, be less opaque to high energy x-radiation than to conventional medical x-rays. It was anticipated that this would have several desirable effects. It would permit penetration of rib shadows to obviate the problem of ribs obscuring pulmonary lesions; it would reduce the contrast of the shadows of ribs and calcified lesions and thus reduce their effectiveness as background distraction; it would reduce the overall film contrast (between blackest black and whitest white) and thus make it possible to record both the lungs and the mediastinum on a single film at densities corresponding to the straight line portion of the film characteristic curve; and, coincidentally, it would provide exceptionally great exposure latitude.

Studies with an anthropomorphic phantom, designed and built by John Harris (5) for dosimery and the evaluation of recording techniques, suggested that, within the then-available kilovoltage range, it was not possible to produce radiographs of significantly different image perceptibility characteristics. Accordingly, it was decided to study the effectiveness in human chest radiography of radiation generated at 1 to 2 million volts--well beyond the conventional diagnostic range. Phantom studies were made with a betatron, a high voltage, condenser-discharge unit designed for ballistic studies, as well as with 1 and 2 MeV resonance transformer and electrostatic generators designed for supervoltage radiation therapy. Ultimately, with the aid of the late Dale Trout and his associates at the General Electric Company, we exposed our first million-volt human chest films (6).

Pilot clinical trials of the "film sandwiching" technique confirmed that "film sandwiching" was feasible and did provide more information than a single chest radiograph (7). It offered nothing, however, that could not be achieved more conveniently by exposing two films independently to different density levels, except that it saved the patient some radiation exposure. Since the very small added exposure to the patient was not then a matter of great concern, this approach was not pursued further.

Early clinical studies of megavoltage radiographs, however, showed that radiographs exposed with such very high energy radiation did indeed have very different physical characteristics from those exposed at conventional voltages. They also indicated: (1) that details of the mediastinum and of the peripheral portions of the lungs might both be satisfactorily recorded in a single exposure; (2) that details of the mediastinum and tracheobronchial tree were far more apparent in megavoltage than in conventional radiographs; (3) that exposures of the human chest could be made in 1 - 2 seconds--not ideal, but perhaps tolerable for chest radiography; and (4) that exposures could be made with an exposure to the patient of approximately 0.8 r--an unacceptable exposure today, but comparable to the exposure associated with photofluorography, which was then in general use.

As a result of these pilot studies, plans were made for the acquisition of a megavoltage generator to be dedicated to a definitive evaluation of the technique. Two types of generator were considered: a 1 MeV resonance transformer generator with a high output and a nominal focal spot of 10 mm; and a 2 MeV electrostatic generator with a low output and a focal spot which, according to our notes at the time, was rated at 1 - 3 mm. It appeared that it would be necessary to choose betwen the unsharpness of a very large focal spot and the unsharpness of motion resulting from the low output of the electrostatic generator. It was difficult then to predict which form of unsharpness might be more acceptable. As an approach to the problem, comparable radiographs of the Harris phantom, with test objects in place, were exposed with the two types of units to determine just what effect the difference in focal spot size might have. To our great surprise, it was not possible reliably to distinguish the radiographs made with a 10 mm focal spot from those made with a 1 - 3 mm focal spot. This finding seemed scarcely credible at the time, but I believe it was an accurate observation.

The definitive clinical trial was never completed because of the press of other duties. However, a more extensive pilot study was completed, the outcome of which, I believe, is of interest here. Two hundred and forty individuals, the majority having chest pathology of one sort or another, were examined by conventional and megavoltage radiography within a period of a few days. The megavoltage radiographs were produced with an electrostatic radiation therapy generator with the cooperation of Dr. Leonard Stanton and the American Oncologic Hospital. The radiographs produced were shown to three trained observers who were asked simply to report whatever they detected that they believed to be abnormal in each radiograph. The megavoltage radiographs were shown first; after an interval of several weeks the conventional controls were presented. (Thus, any memory of the individual cases would have favored the interpretation of the conventional radiographs.) The readers then, as a group, reviewed both sets of films and all of the reports pertaining to each case and developed a consensus judgment of what findings were, in fact, present. That is to say, they developed a consensus with regard to what should have been reported in each case. An independent reviewer then compared the readers' individual reports with this consensus judgment and findings were scored as "reported" or "missed." No specific record was made of false positives because they were extremely few in number, and ROC analysis had not yet been developed, so the importance of such data was not recognized.

Let us look now at the relative frequency with which findings were perceived in the two types of radiographic recordings as a function of the nature of the abnormality.

Site or Type of Lesion	Reported		Missed	
	80 kVp	2 MeV	80 kVp	2 MeV
Tracheobronchial Tree, Mediastinum, Abnormal aeration	59	91	67	35
Lungs, Pleura	326	330	109	105
Heart, Vessels, Hila	68	48	28	48
Ribs, Granulomata, Chest Wall	26	13	40	53

These data would appear to confirm the previous impressions relative to the value of supervoltage radiography. The number of mediastinal abnormalities and abnormalities of aeration reported in the megavoltage radiographs was, statistically, significantly greater than the number reported in conventional radiographs; the number of skeletal lesions and other calcified abnormalities reported in the megavoltage radiographs was significantly smaller than the number reported in the conventional radiographs; and there was no significant difference in the number of pulmonary parenchymal and pleural abnormalities reported.

Clearly, improved mediastinal recording was achieved in supervoltage radiographs at the expense of the recording of calcified details. Depending on the nature of the clinical problem, this might or might not offer diagnostic advantage. Also, it should be noted that the expectation that peripheral nodules would be more likely to be perceived in the megavoltage radiographs then in the conventional studies was not realized. This may have been an artifact of patient selection, or it may have been an artifact of the reporting system employed; whatever the explanation, peripheral nodules were perceived no more often in megavoltage films than in conventional films.

Now, let us look at the data presented in a somewhat different manner. Here, only the total numbers of abnormalities "reported" and "missed" in the two types of recordings are compared.

	Reported	Missed
Supervoltage Radiographs	482	241
Conventional Radiographs	479	244

Note that, despite the dramatic difference in the physical characteristics of the two types of recordings, there was virtually no difference in the total number of findings reported, nor was there any difference in the fraction of the recorded findings "missed," i.e., in the percent of false negatives. It is as if the data reflected some fundamental attribute of the observers, rather than differences in two types of radiographic recording.

Now, what are the "lessons" to be learned from this experience? First, I believe that the experience cited suggests that chest radiography has evolved to a point at which technical modifications designed to improve the recording of one type of tissue or structure will almost certainly have an adverse effect on the recording of some other type of tissue. This simply reflects the fact that the imaging problems associated with various types of tissue within the thorax vary tremendously, and the solutions to these problems may well be mutually exclusive. I would suggest, therefore, that there probably is no single optimum radiographic technique for the chest. Rather, there are a variety of more or less adequate compromise techniques. The question is not, "What one technique is best?" but, rather, "Which technique represents the most favorable compromise given this particular clinical problem?"

Second, I believe the experience reported supports the view that some fraction of the clinically significant findings recorded in the radiograph will inevitably escape the notice of the radiologist. This fraction probably varies with the anatomic nature of the study but, with respect to chest radiographs, is of the order of 20 - 30 percent. Gross overexposure or underexposure of the radiograph, gross unsharpness, or significant degradation of contrast may increase this fraction. I believe, however, that the most meticulous of techniques is unlikely to reduce it significantly below 20 percent because, within broad limits, it is not determined by the "quality" of the radiograph, but rather by the mechanisms of human perception. This I submit raises the fundamental question of whether there is more to be gained by seeking to optimize the radiographic process or by seeking to develop economical systems to provide for replicate reading--the only device so far shown significantly to improve retrieval of recorded diagnostic information.

Third, our experience suggests that it is exceedingly difficult to understand clinical imaging problems by studying isolated elements of such complex absorbers as the human chest. The interactions of the many shadows of the chest radiograph significantly limit the perception of critical shadows. Studying phantom lesions in isolation, in the absence of such interactions, has little bearing on the problems of clinical radiographic interpretation.

Fourth, I believe that our inability to appreciate the effects of focal spot unsharpness in supervoltage radiographs indicates that whereas technical forms of unsharpness (e.g., focal spot size, film grain size) limit the recording of high frequency, high contrast, closely approximated details such as those of trabecular bone, the unsharpness of the shadows of most soft tissue strucures and lesions is determined by subject unsharpness, i.e., unsharpness determined by the contour of the absorber. Hence, when we suppressed bone detail by using megavoltage radialton, the effect of changing focal spot size was not apparent. This observation has a very interesting theoretical implication. If the signal with which we are concerned is a gross soft tissue shadow (and most masses, consolidations, infiltrates, and the like produce such shadows); if we think of superimposed ribs as a form of noise; and if we accept the somewhat shakier premise that the "sharper" a shadow the more demanding it is of attention; then efforts to reduce technical unsharpness have no effect on the signal, but may make this form of noise more demanding of attention. In short, I submit that in seeking "technically optimized" radiographs there is at least a theoretical possibility that we are in fact adversely affecting the signal-to-noise ratio with respect to the recognition of most pulmonary lesions.

In summary, for a period of 10 years between 1952 and 1962, we studied the potential application of magavoltage radiation in clinical chest radiography. We demonstrated, I believe, that such radiography was technically, if not economically, feasible. We demonstrated that, relative to soft tissue, calcified structures are less opaque to megavoltage radiation than to conventional diagnostic x rays, and that this difference does permit improved recording of mediastinal details without concomitant overexposure of the lungs. We found that trained observers did perceive different types of abnormalities in megavoltage and conventional radiographs--the perception of more mediastinal details in megavoltage studies being achieved at the cost of perceiving fewer abnormal calcifications and skeletal abnormalities. Since different types of findings are of importance in different clinical circumstances, this would suggest that, at least in selected problems, megavoltage radiography would be more effective than conventional despite the fact that the total number of discrete abnormalities identified and the percentage of false negative reports was the same in the interpretation of megavoltage radiographs as in the interpretation of conventional radiographs. Most significantly, perhaps, we found that the radiologic community did not rush to accept the concept of megavoltage radiography--possibly because of the technical limitations of the equipment we employed; more probably, I should judge, because the perceived advantages did not suffice to offset the cost and loss of skeletal detail. In any event, it is heartening to hear the subject of high voltage (even if not megavoltage) radiography revived in a conference such as this; perhaps the concept will yet prove useful in clinical chest radiography.

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THE PHYSICAL BASIS FOR HIGH VOLTAGE CHEST RADIOGRAPHY

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INTRODUCTION

Overall visibility in chest x-ray films improves at high voltage. Improvements have been reported over a wide range of voltages: 150 kVp in combination with a special tunnel wedge filter (1), 200 kVp (2), 250 kVp (3), 350 kVp (4, 5), 1000-2000 kVp (6-8).

The improvement is substantial: bone shadows are reduced, making soft tissues more visible behind ribs, at lung peripheries and in the mediastinum. Visualization of pulmonary nodules and vasculature is enhanced. Depth resolution is increased in the thicker or bony portions of the chest, causing thinner tissues to cast a visible shadow (9). Figure 1 illustrates some of these features. In 350 kVp films (compared to 120 kVp films) significant improvements are reported in visibility and diagnosis of the infradiaphragmatic and retrocardiac areas, the larynx, trachea and mediastinum, cardiac abnormality, pleural disease, interstitial and alveolar diseases (5).

In this paper we analyze the characteristics of low and high voltage chest films. The analysis is based on the concepts of film latitude and contrast, matching of radiological image to film response, depth resolution and total resolution, and conspicuity. Two main conclusions are reached:

First, the visibility of small features in the chest is generally limited much more severely by depth resolution (1-20 mm) than by lateral resolution (0.2 to 0.6 mm); hence, source size is relatively unimportant, within reason, and the largest opportunity for better visibility lies in improved depth resolution, a conclusion supported by Harris (3), Tuddenham (10) and Thomas (2).

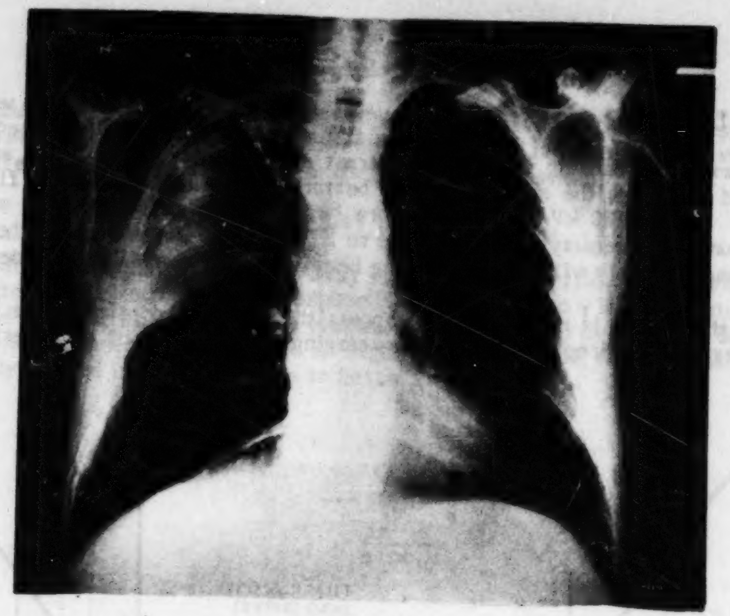
Second, the most serious limitation to depth resolution in conventional chest films occurs because a large portion of the image is obscured by bone and recorded at low film contrast; a high voltage technique is more effective than a low voltage technique in optimizing overall depth resolution.

The two films in Figure 1 illustrate typical differences between PA chest films at low and high voltage. The 90 kV film has greater contrast and sharpness; however, large portions of the chest are either overexposed or underexposed, yielding little information. The 350 kV film has lower contrast and sharpness--shades of gray rather than blacks and whites--yielding more information.

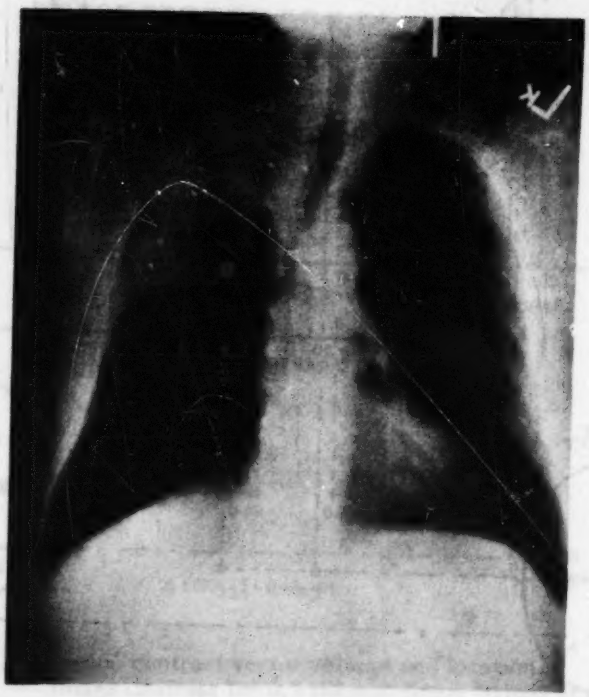
X-ray attenuation decreases with increased voltage in both tissue and bone. The decrease is relatively slow in tissue (20 percent from 120 kVp to 350 kVp) and much more rapid in bone (40 percent). The transmitted x-ray beam is imaged on film. Film characteristics and limitations have a strong effect on visibility.

X-RAY IMAGE AND FILM LATITUDE

Figure 2 illustrates schematically the response of a typical x-ray film (H & D curve for Kodak XRP film) and the exposure through two extreme paths in a simplified chest model: minimum absorption path through the lungs and maximum absorption path through the mediastinum. Film exposure is set for a typical maximum lung field density of



(a) 90 kVp



(b) 350 kVp

Fig. 1. Chest films of the same patient.

approximately 1.8 (B). At 120 kV, the minimum density in the mediastinum is very low (A); the latitude of the x-ray image exceeds the latitude of the film and the mediastinum is strongly underexposed, causing very poor contrast and visibility. At 350 kV and with the same lung field exposure, the mediastinum is better exposed (A'), increasing film contrast and visibility. If the 120 kV film were more heavily exposed to enhance mediastinum visibility, the lung field density would increase to 2.7 (B'). The lungs would be be burned out and could be examined only with bright-lighting because of the excessive film density range.

Hence a high voltage is required to compress the range of density in the radiological image to the extent required for acceptable imaging of all points of the chest in a single film.

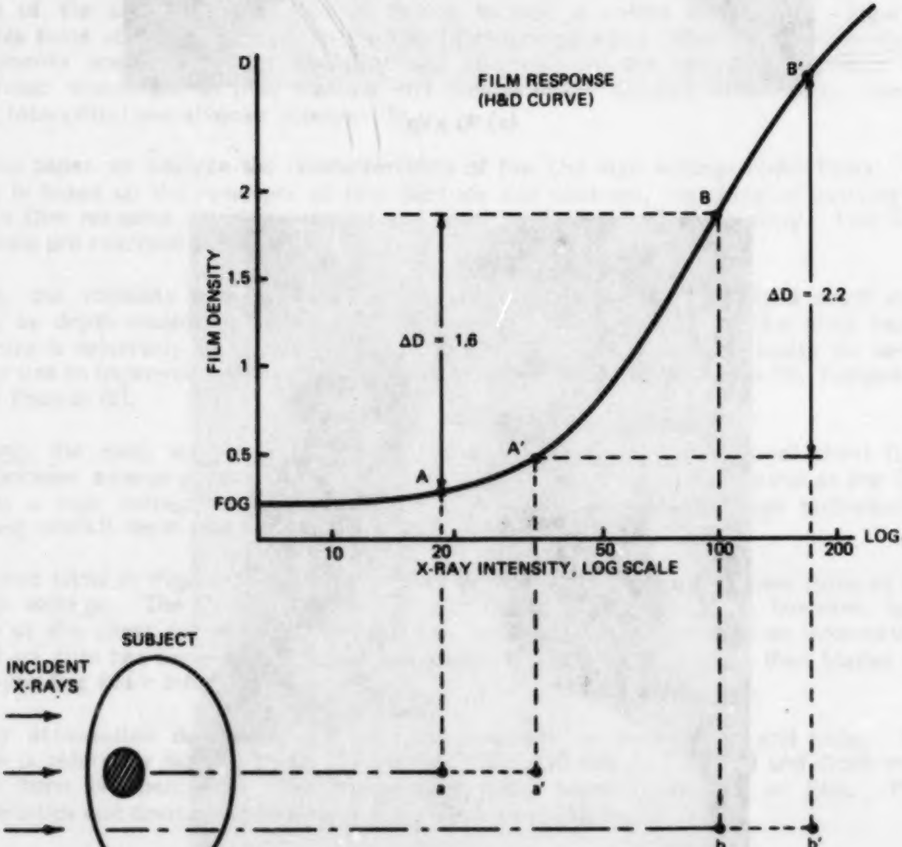


Fig. 2. Film exposure versus voltage.
Proper exposure at low voltage AB.
Proper exposure at high voltage A'B.
Overexposure at low voltage A'B'.

DEPTH RESOLUTION

The fundamental film contrast relation and the relative visibility of lesions or nodules are considered in Figure 3. A nodule can only be seen if it casts a sufficient shadow ΔD. Assuming effective grids, ΔD is given by the familiar expression:

$$\Delta D = 0.43\ \mu \gamma \Delta x$$

hence ΔD depends equally on the X-ray attenuation coefficient μ and on the film contrast factor γ (9). In the properly exposed lung field where γ is maximum, film contrast will decrease with increasing voltage. However, as illustrated in Figure 3, film contrast in the mediastinum can actually increase at high voltage because the decrease in μ is more than offset by a large increase in γ due to better film exposure.

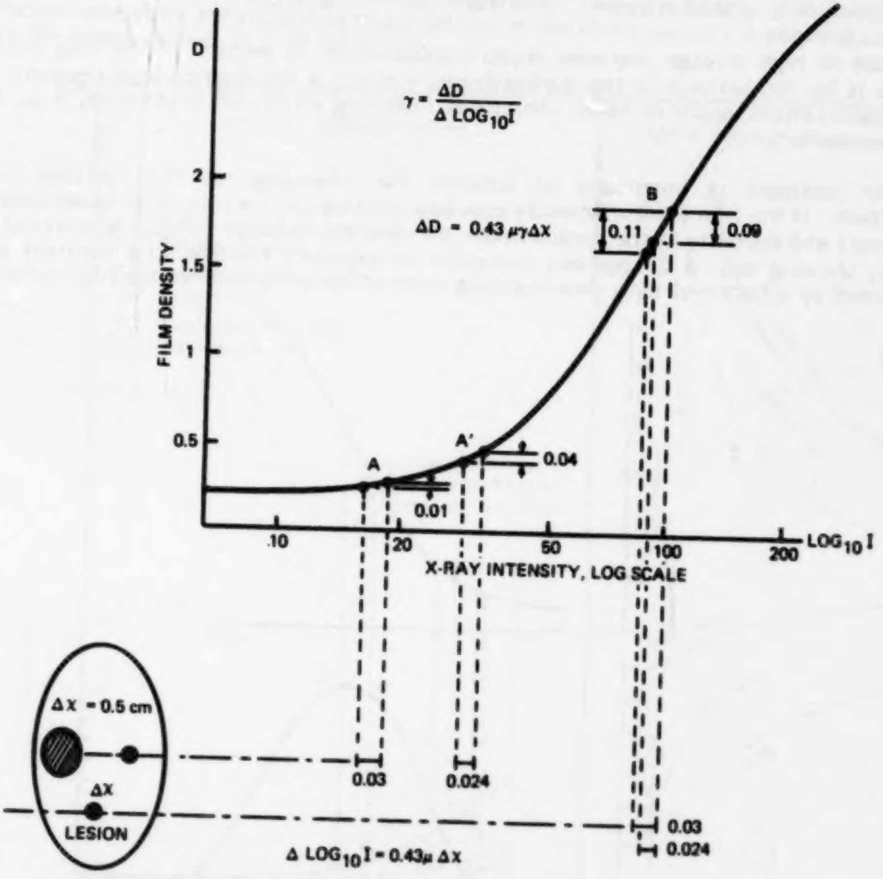


Fig. 3. Lesion contrast versus voltage and location. Image contrast for 0.5 cm soft tissue lesions in the lung field and in the mediastinum, at 350 and 120 kV.

In the specific example of Figure 3, a .5 cm soft tissue lesion reduces the transmitted x-ray intensity by 7 percent at 120 kV and 5.8 percent at 350 kV. The resulting $\Delta \log_{10} I$ equals 0.03 and 0.024, respectively. If the lesion is in the lung field where the film contrast factor is maximum ($\gamma = 3.6$ at B) the resulting shadow has a contrast $\Delta D = 0.11$ at 120 kV and 0.09 at 350 kV. However, for a similar lesion located in the mediastinum, the film contrast at 120 kV is reduced to $\gamma = 0.33$ (A) due to severe underexposure and the lesion image has a contrast of only $\Delta D = 0.33 \times 0.03 = 0.01$. At 350 kV the mediastinum is better exposed (A'), γ is increased to 1.7, and the lesion image contrast increases fourfold to $\Delta D = 0.04/0.43\ \mu\gamma$.

The minimum perceptible ΔD depends on a number of factors: nodule size and shape, sharpness of boundaries, obscuring shadows, quantum noise, film exposure and viewing distance. Under favorable conditions, the minimum ΔD is of the order of $0.04/0.43\ \mu\gamma$ and the depth resulution, i.e., the minimum perceptible soft tissue nodule thickness, is given by

$$\Delta x_{min} = \frac{0.04}{0.43\ \mu\gamma}$$

	ΔX_{min}	
	120 kV	350 kV
Lung field	0.17 cm	0.22 cm
Mediastinum	2.0 cm	0.5 cm

Hence, use of high voltage degrades depth resolution by 30 percent in the lung field but improves it by 300 percent in the mediastinum, yielding a net overall improvement. The same considerations apply to other structures involving air/tissue boundaries, e.g., blood vessels, airways.

Proper exposure is important to achieve the advantage of high voltage in the mediastinum. If the film is insufficiently exposed, contrast in the lung field is not improved but contrast and visibility in the mediastinum are severely reduced. This is illustrated in Figure 4, showing that a 25 percent reduction in exposure reduces film contrast in the mediastinum by a factor of two, thus negating most of the advantage of the high voltage.

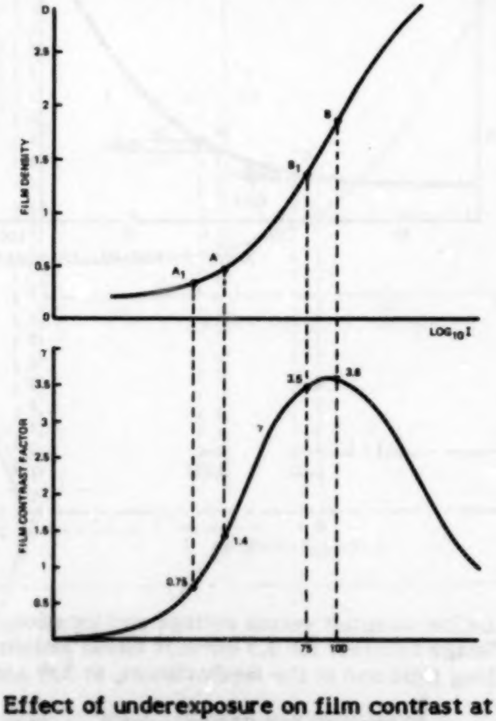


Fig. 4. Effect of underexposure on film contrast at high voltage: proper exposure range AB; insufficient exposure A_1 B_1. Film contrast in the mediastinum is reduced by a factor of two.

TOTAL RESOLUTION

Resolution, which measures the minimum dimensions an object must have to be imaged on film, has two components: lateral resolution Δy, associated with the unsharpness of the x-ray imaging system, determines the minimum visible width; depth resolution Δx determines the minimum visible thickness. It is the larger of these two components which controls system total resolution and which must be minimized to enhance visibility of small detail.

These factors can be calculated and compared, as shown in Figure 5. The lateral resolution calculations are for a 350 kVp field emission system with an equivalent 2-mm focal spot, and a conventional system with a 1-mm nominal focal spot. The important result is that depth resolution is much poorer than lateral resolution, even for bone or calcifications. Thus, depth resolution dominates visibility, and focal spot size is not a significant factor in soft tissue or nodule visibility. This conclusion is supported by the study of detection accuracy in chest radiography by Gray et al., who found that the detection accuracy for simulated tissue lesions in the chest was not dependent on focal size over the range studied (0.3 - 2 mm) (11).

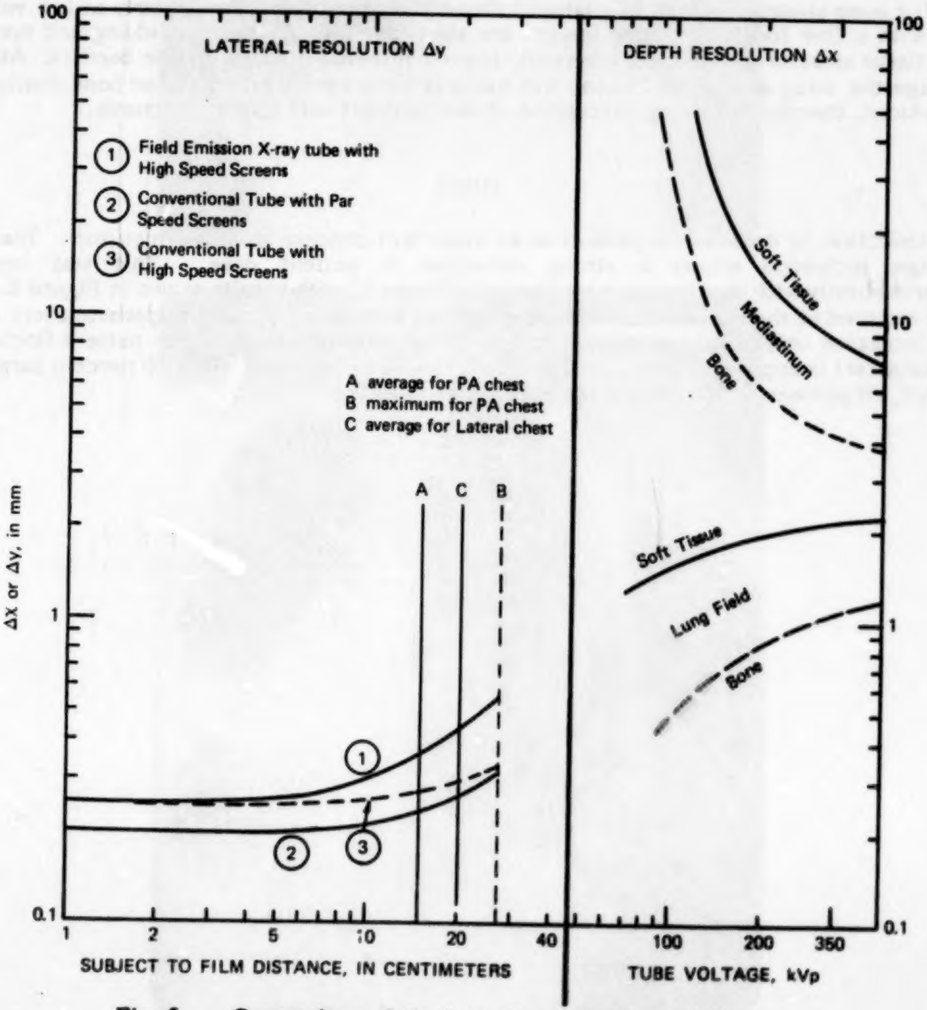


Fig. 5. Comparison of the lateral and depth resolution limitations to visibility of small detail.

CONSPICUITY

The increased visibility of vascular markings and soft tisue lesions in high voltage chest films is often much greater than can be explained solely by physical analysis of image contrast and depth resolution (9). The added visibility seems due to enhanced pattern recognition: bone shadows are less intrusive and the contours of diffuse low contrast soft tissue patterns are more easily recognized. This agrees with earlier observations by Tuddenham (10) and Morgan (12). Kundel has recently presented evidence of the detrimental effect of "structured noise" which inhibits the perception of low contrast lesions (13, 14). A pair of 350 kV bronchograms of the same patient, shown in Figure 6, provides a striking demonstration of conspicuity, i.e., the ability of a sharp high contrast background to inhibit perception of lower contrast underlying structures. The two films were taken with the same 350 kV system and the exposures are closely matched. In the first film the contrast medium failed to penetrate the right lung and all ribs are clearly visible. In the second film the contrast medium provides a sharp, high contrast bronchial pattern which obliterates the ribs from view.

The same situation occurs in relation to bone and soft tissue. Particularly at low voltage and with a fine focal spot, bone images are sharp and high contrast, subduing and masking soft tissue structures with their inherently lower contrast and more diffuse borders. At high voltage the x-ray absorption in bone and tissue is more evenly matched and bone dominance is reduced, thereby enhancing perception of low contrast soft tissue structures.

DOSE

Reduction of dose to the patient is an important concern of this symposium. The high voltage technique allows a strong reduction in patient dose. This was verified experimentally with the arrangement shown in Figure 7, with results shown in Figure 8. The dose required at the cassette for a fixed lung field exposure (D = 1.8) is approximately 1 mR and increases only slowly at high voltage. Total entrance dose to the patient (including backscatter) is approximately 15 mR in the 250 - 350 kVp range. It is 50 percent larger at 150 kV, 70 percent at 120 kV and 100 percent at 100 kV.

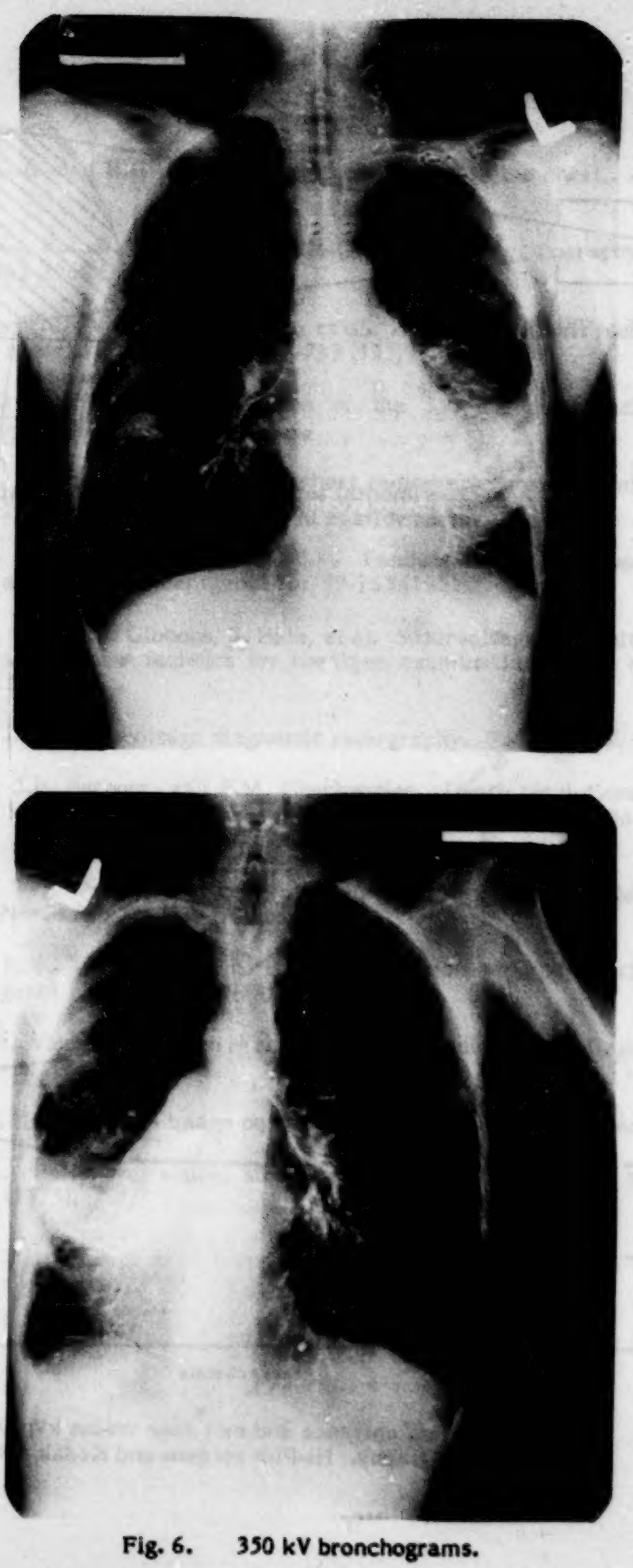


Fig. 6. 350 kV bronchograms.

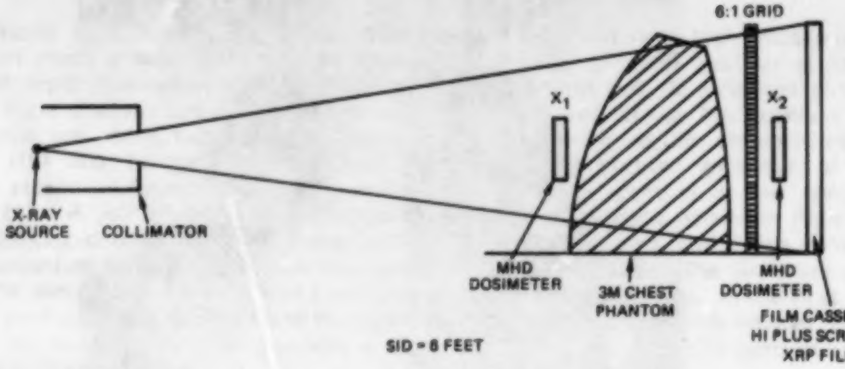

6:1 GRID
X_1
X_2
X-RAY SOURCE
COLLIMATOR
MHD DOSIMETER
3M CHEST PHANTOM
MHD DOSIMETER
FILM CASSETTE
HI PLUS SCREENS
XRP FILM
SID = 6 FEET

Fig. 7. Experimental setup for measurement of patient dose versus voltage in chest radiography.

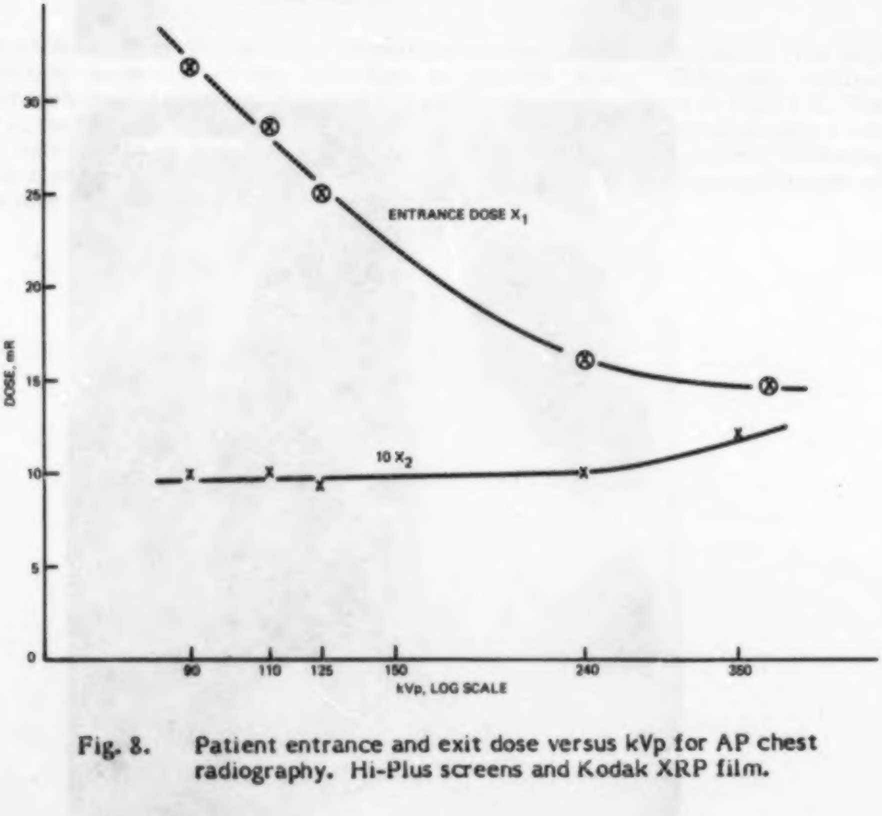


Fig. 8. Patient entrance and exit dose versus kVp for AP chest radiography. Hi-Plus screens and Kodak XRP film.

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A COMPARATIVE ANALYSIS OF 350 kVp AND 120 kVp CHEST RADIOGRAPHY

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INTRODUCTION

The recent interest in high kilovoltage chest radiography is not a new phenomenon (1-4). Portal films exposed with cobalt teletherapy sources and with other supervoltage equipment have been used for many years in radiation treatment planning. These films exhibit the characteristic decrease in image contrast related to the marked decrease in the mass absorption coefficient of bone relative to the tube potentials generally used in chest radiography (3).

Applying the advantages of the radiotherapy portal film to diagnostic chest radiology, Tuddenham et al. produced chest radiographs exposed in the range of 1,000-2,000 kVp. Although the exposure times were long and the exposure to the patient was relatively high, the advantages of simultaneous visualization of the mediastinum and the lung fields, as well as the improved visualization of the lung periphery because of low contrast of bony structures, suggested that the high kilovoltage technique might offer certain advantages (2,4).

Subsequently, experimental studies using the field emission x-ray tubes suggested that a voltage in the 250-400 kVp range might be optimal for chest radiographic technique by combining some of the advantages of both high kilovoltage and low kilovoltage techniques resulting from its unique energy spectrum (5,6).

In this range, conventional film/screen combinations and grids can be used, and the problems of higher patient dose and long exposure times are reduced.

In 1971, a 300 kVp prototype diagnostic radiographic machine was tested and, subsequently, a production model 350 kVp system was built by the Hewlett-Packard Corporation and tested in several large centers (5-8).

Dyke, Charbonnier, and others explored the physical basis for improved image visibility in the 300-400 kVp range. This material was discussed in depth earlier in the symposium and in several technical papers published by the Hewlett-Packard Corporation (5-7, 9,10).

Proto and Lane in 1978 compared 120 and 350 kV techniques using different film/screen combinations for each of the compared studies. They performed their examinations in separate rooms. Their results suggest that the higher kilovoltage technique was superior in all areas except in bone and in the production of mottle (11).

In order to reduce variation to a minimum and thus yield statistically more valuable results in a comparative study, we again compared the 120 and 350 kVp techniques following standardization of technical factors and installation of both systems in a single room using the same automatic film changer.

METHODS AND MATERIALS

Both the 350 kVp posterior-anterior and lateral films and the 120 kVp posterior-anterior and lateral films were produced in the same conventional radiographic room in the outpatient department of the Thomas Jefferson University Hospital during the first six months of 1978.

The examinations were all performed at a tube-object distance of 1.8 meters using DuPont SP screens and Agfa Gevaert RP-L film. A Leibel-Flarsheim 12:1 ratio, 103-line focused stationary grid was inserted in a DuPont automatic film changer and used for both techniques. The films were processed in a Kodak M7 120-second processor at a temperature of 93 °F. PXR developer and fixative were used.

The 350 kVp radiographs were produced with a Hewlett-Packard Model 43815 x-ray machine. A fixed 40 mA current was used. The average exposure time was 10 milliseconds. The half-value layer for the 350 kVp system was 12.5 mm of aluminum and 1.4 mm of copper. The measured focal spot diameter was 2.5 mm using a star resolution test object.

The 120 kVp studies were exposed using a standard 600 mA single-phase generator. The thermionic tube had a 2 x 3.1 mm measured focal spot. The average exposure time was 10-12 milliseconds. Automatic exposure phototiming was used with both systems.

The recommended density factors in the lung fields in the range of 1.4 to 1.8 were followed initially. During the study, a reduction in density to 1.1 was noted during routine maintenance of the equipment and density was subsequently corrected to 1.6 .

Physical measurements of phototimed exposures showed that the average exposure with a 12 cm water-equivalent phantom was 14 milliroentgens at 120 kVp. The average exposure of the 350 kVp system was 9 milliroentgens.

The thermionic and field emission tubes were mounted one above the other. The 120 kVp tube housing was placed such that the central axis of the primary beam was at a 7° angle in relation to the primary beam of the 350 kVp tube. The 350 kVp tube was placed in a horizontal position with the axis of the primary beam at 90 degrees from the surface of the cassette. The primary beams of both tubes were centered on the fixed screens and the patient was positioned with an elevator.

COMPARATIVE EVALUATION OF 120 AND 350 kVp TECHNIQUE

The PA and lateral 350 and 120 kVp exposures of 430 patients were compared by five experienced radiologists. The 350 kVp and 120 kVp films were read separately, and a computer-compatible analysis form was completed by the reader. The two studies to be compared were read one week apart in an attempt to eliminate any memory of the patient's films and thus reduce observer prejudice.

Each reader was requested to evaluate 21 normal anatomical landmarks divided among soft tissue and bony structures, pulmonary parenchyma, mediastinal and pleural anatomic areas. Decisions on the presence or absence of six pathologic patterns were also requested from the readers.

In order to simplify the study and to rapidly accumulate statistically significant information, the presence or absence of normal anatomical structures was emphasized. It was assumed that visibility of normal landmarks would also suggest visibility of abnormal anatomy in the same region.

Five levels of visibility were used for evaluation, from the lowest level of "structure not seen" to the highest level of "structure well seen." The reader also indicated whether or not high-density illumination had been used. Comments concerning noise level and exposure technique were also requested.

Following completion of the cases presented to the readers, the results were submitted for statistical analysis using the chi square probability method.

RESULTS

The results of the first 330 cases were compared as presented in Table 1. Table 2 presents a comparison of an additional 100 cases following an adjustment in the 350 kVp phototiming density factor to 1.6 . The results from the initial comparison show that there was better visualization of the descending aorta, the carina, the posterior junctional line, and the aortico-pulmonary window to a statistically significant level of probability using the 350 kVp technique. No difference was observed in the identification of the right border of the left atrium. This was due mainly to the large number of studies in which the left atrium was not identified during the examination in either study.

Table 1. Comparison of 350 and 120 kVp chest radiographic technique stressing normal anatomic landmarks[+]

	350 Better		120 Better		No Difference	Total No. of Patients
	%	Signif.	%	Signif.	%	
MEDIASTINUM	POSTERIOR-ANTERIOR PROJECTION					
Descending Thoracic Aorta	35	*	25	---	40	330
Rt. Border of Lt. Atrium	23	N.S.	19	---	58	329
Carina	42	**	18	---	32	331
Posterior Junctional Line	34	**	19	---	47	309
Aortico-Pulmonary Window	30	*	21	---	49	324
LUNG						
Lt. Lung Apex	28	---	32	N.S.	40	326
Rtrocardiac Vessels	22	---	36	**	42	319
Pulmonary Vessels in Rt. Mid Lung Field						
- inner half	20	---	28	*	52	328
- outer half	24	---	35	*	41	308
Lt. Vertebral-Phrenic Angle	38	**	19	---	43	305
Minor Fissure	22	---	37	**	40	326
BONES AND SOFT TISSUES						
Dorsal Spine below Carina	18	---	27	*	55	329
Rt. Anterior Ribs	17	---	69	**	14	328
Axillary Soft Tissues	16	---	53	**	31	328
Supraclavicular Fossa	16	---	46	**	38	324
Subdiaphragmatic Soft Tissues	21	---	35	**	44	327
	LATERAL PROJECTION					
Inferior Vena Cava	22	---	33	*	45	199
Aortico-Pulmonary Window	29	N.S.	25	---	46	170
Epicardial Fat	18	---	21	N.S.	61	196
Lt. Pulmonary Artery	34	*	23	---	43	198
Major Fissure	17	---	39	**	39	198
Lung Apices	46	**	18	---	37	200

[+] evaluated for statistical significance by the Chi Square Probability Test

*$p < .05 > .01$

**$p < .001$

N.S. = $p > .05$

Table 2. Additional series of 100 patients with higher phototimer density factor: Comparison of 350 and 120 kVp chest radiographic technique stressing normal anatomic landmarks+

	350 Better %	 Signif.	120 Better %	 Signif.	No Difference %
	POSTERIOR-ANTERIOR PROJECTION				
MEDIASTINUM					
Descending Thoracic Aorta	84	**	3	---	13
Rt. Border of Lt. Atrium	13	N.S.	8	---	79
Carina	94	**	0	---	6
Posterior Junctional Line	66	**	13	---	31
Aortico-Pulmonary Window	63	**	13	---	34
LUNG					
Lt. Lung Apex	41	*	13	---	46
Retrocardiac Vessels	46	**	8	---	46
Pulmonary Vessels in Rt. Mid Lung Field					
- inner half	8	---	38	**	54
- outer half	8	---	49	**	43
Lt. Vertebral-Phrenic Angle	76	**	3	---	22
Minor Fissure	13	---	39	*	48
BONES AND SOFT TISSUES					
Dorsal Spine below Carina	61	**	0	---	39
Rt. Anterior Ribs	6	---	84	**	10
Axillary Soft Tissues	8	---	64	**	28
Supraclavicular Fossa	8	---	72	**	20
Subdiaphragmatic Soft Tissues	54	**	3	---	43
	LATERAL PROJECTION				
Inferior Vena Cava	24	---	39	N.S.	37
Aortico-Pulmonary Window	30	N.S.	24	---	36
Epicardial Fat	33	N.S.	20	---	47
Lt. Pulmonary Artery	53	*	24	---	24
Major Fissure	22	---	47	*	20
Lung Apices	77	**	20	---	3

+evaluated for statistical significance by the Chi Square Probability Test

*$p < .01 > .001$

**$p < .001$

N.S. = $p > .05$

On the other hand, with the 120 kVp technique there was better visualization of most anatomic parts related directly to the pulmonary parenchyma. There was no statistical difference in appreciation of detail in the left lung apex. However, the 350 kVp technique was statistically superior in the left vertebral phrenic angle. Soft tissue landmarks were better seen by the 120 kVp technique, as were the anterior ribs and the dorsal spine.

In the lateral view, the 350 kVp technique was superior in the lung apices and in the hilum. On the other hand, the epicardial fat and the aortico-pulmonary window were visualized equally well with either 120 or 350 kVp technique. The inferior vena cava and major fissure were better seen with the 120 kVp technique in the lateral projection.

As several anatomic areas were better seen with the lower kilovoltage technique, where one could theoretically expect the 350 kVp technique to be superior, our exposure factors and equipment were re-examined in an attempt to explain this unexpected result. Our review suggested that an upward readjustment of the 350 kVp phototiming density factor to a level of 1.6 would be advantageous. Following this, an additional 100 cases were evaluated. Results using the same anatomic parameters showed a marked improvement in visualization with the 350 kVp technique in those areas questioned in the initial evaluation.

These included the dorsal spine, where in 61 percent of the cases there was better visualization with the 350 kVp technique. The infradiaphragmatic soft tissues and, in the lateral projection, the aortico-pulmonary window, epicardial fat, left pulmonary artery and lung apex, were also better seen with the higher kilovoltage method.

In this additional series of patients, the 350 kVp technique was superior in the mediastinum in all categories. The 350 kVp technique was superior in the lung apex, because of reduction of bone visualization in this area. In the retrocardiac area, the arteries and veins were seen better using the 350 kVp technique in 46 percent of the patients, compared to 8 percent using the 120 kVp technique. However, visualization of the peripheral pulmonary vasculature was superior using the 120 kVp technique.

There was general agreement that the 350 kVp technique demonstrated a higher level of noise (mottle). Because of the small number of pathologic entities evaluated, no tabulation of the pathologic patterns was done.

DISCUSSION

Prior to the discussion of the results of our study let us review for a moment the several reasons that have been suggested to explain the rationale for improved visualization of anatomical detail using the 350 kVp technique. These include:

1. A decrease in the ratio of the mass absorption coefficients of bone and soft tissues.
2. Differential visibility of bone detail in the mediastinum.
3. An increase in depth resolution.
4. Higher conspicuity of potentially significant shadows relative to structured noise.

1. A decrease in the ratio of the mass absorption coefficient of bone and soft tissue occurs as kilovoltage increases above the conventional levels. At 90 kVp the mass absorption coefficient of bone is greater than that of soft tissue by a factor of about 4.5 . For this reason, bone images are well seen relative to soft tissue. The absorption coefficient of soft tissue remains relatively constant, decreasing only 17 percent from 150 to 350 kVp, while the absorption coefficient of bone is greatly reduced at 350 kVp. Consequently, at 350 kVp, the ratio between the mass absorption coefficients of bone and soft tissue decreases to 1.5. The result of this decrease is better visibility of the lung fields and mediastinum because of less interference from bony structures.

2. Differential visibility of bone detail. Bone detail is better seen in the mediastinum at 350 kVp than at 120 kVp. On the other hand, significant bone detail is lost in the periphery of the thorax. This seemingly paradoxical situation also applies to calcifications. This improvement of visualization of bone and calcifications with the 350 kVp technique in the mediastinum is thought to be due to increased depth resolution by virtue of better penetration.

3. An increase in depth resolution. Depth resolution is a measure of the thinnest tissue that will cast a detectable x-ray shadow. Depth resolution consequently is a limiting factor in visibility. The smallest soft tissue lesion that will cast a detectable shadow is about 1 mm in diameter. In overexposed and underexposed areas of the film, depth resolution may be degraded by a factor of up to 20. Since overexposure and underexposure can be more easily avoided in the higher kilovoltage range, depth resolution usually is not so degraded. Depth resolution of soft tissue increases substantially in the mediastinum, an

area usually underexposed in conventional films. For example, at 150 kVp the smallest mediastinal lesion that generally can be seen is 2 cm in diameter. The limit of visibility is 5 mm in diameter at 350 kVp. In the open lung field, depth resolution decreases with kilovoltage, although most of the decrease occurs below 150 kVp. An additional decrease in depth resolution of about 17 percent occurs between 150 and 350 kVp.

4. Higher conspicuity. The results of this study are at variance with those of Proto and Lane, Hallenbach, and others (8,11,13). While the previous investigators emphasize the improvement in visualization of lung detail, including pulmonary nodules, infiltrates and vascular structures, our evaluation strongly suggests that there is indeed a degradation of detail in the lung periphery. This degradation is thought to be due to a reduction in depth resolution in the lung periphery as well as a higher conspicuity factor. In other words, a relative increase in information occurs in the higher kilovoltage range together with decreased depth resolution, resulting in a decreased ability to define abnormalities in an anatomically "busy" background (12-14).

In this comparative evaluation, visualization in the lateral view is found to be strikingly superior using the 350 kVp technique. This has resulted in our current use of the combination of 120 kVp PA films and 350 kVp lateral films as a standard combination in our outpatient department.

The results of our study lead us to suggest that the 350 kVp film should not serve as a primary study but rather should be used extensively in a tailored chest examination. Our impression at this time is that the 350 kVp study is most valuable when:

1. Improved penetration and fine detail in the mediastinum are particularly important;
2. Soft tissue densities overlie the lung fields, obscuring information, e.g., soft tissues in heavy patients or women with large or dense breasts;
3. Postoperative followup patients have mediastinal, cardiac, or pulmonary pathology; and
4. Patients have heterogeneous aeration, e.g., emphysema, cystic or bullous disease, or have extensive or diffuse pulmonary parenchymal disease.

Because it is relatively simple for the technician to perform either the 350 kVp or 120 kVp study in a dual radiographic room, we tailor the study by making decisions regarding technique immediately before the examination. This may be done by obtaining information on such items as previous surgery, previous films in the department and history of the current complaint. All films should be made available prior to the performance of this study. In this way, we hope to use each technique to its best advantage.

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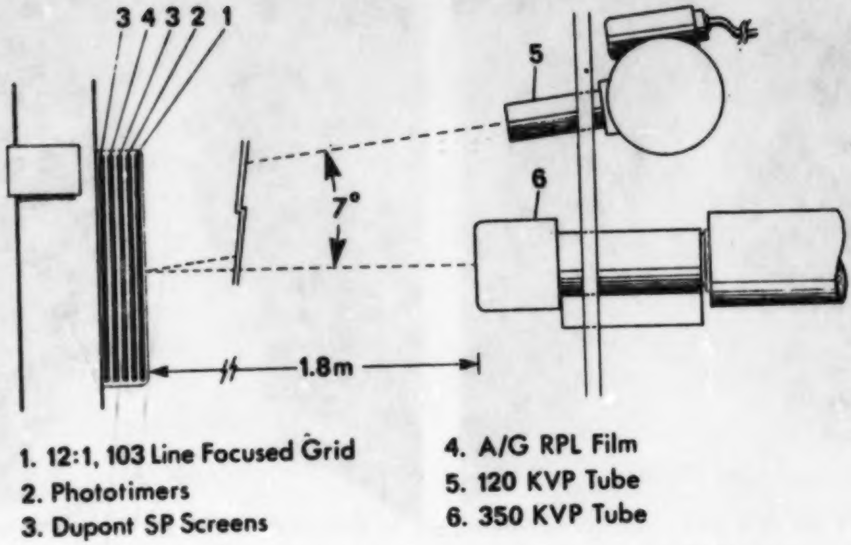
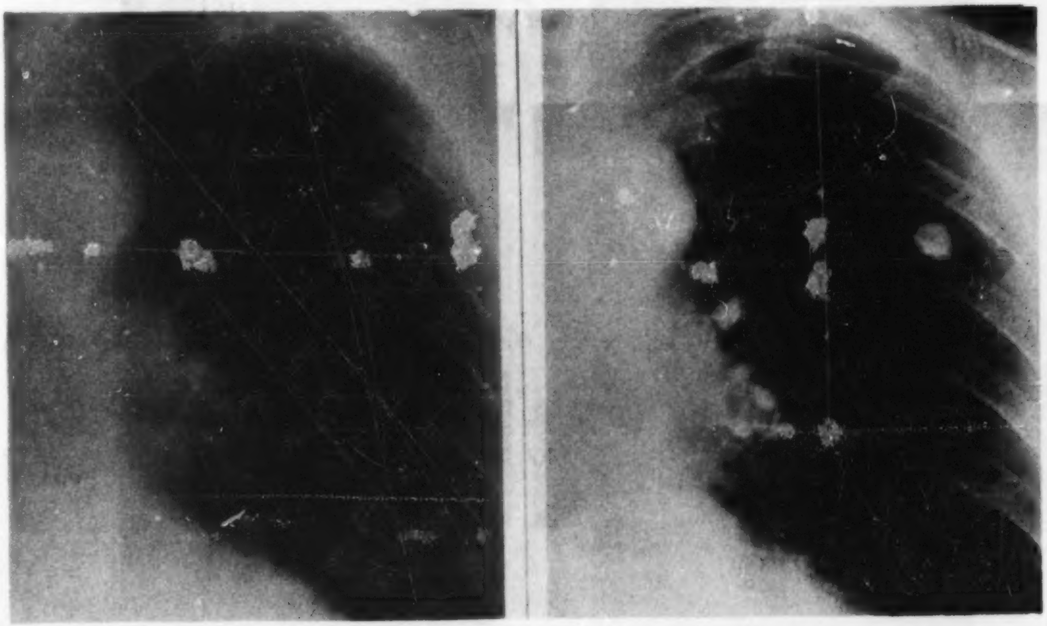

3 4 3 2 1
5
7°
6
1.8m
1. 12:1, 103 Line Focused Grid
2. Phototimers
3. Dupont SP Screens
4. A/G RPL Film
5. 120 KVP Tube
6. 350 KVP Tube

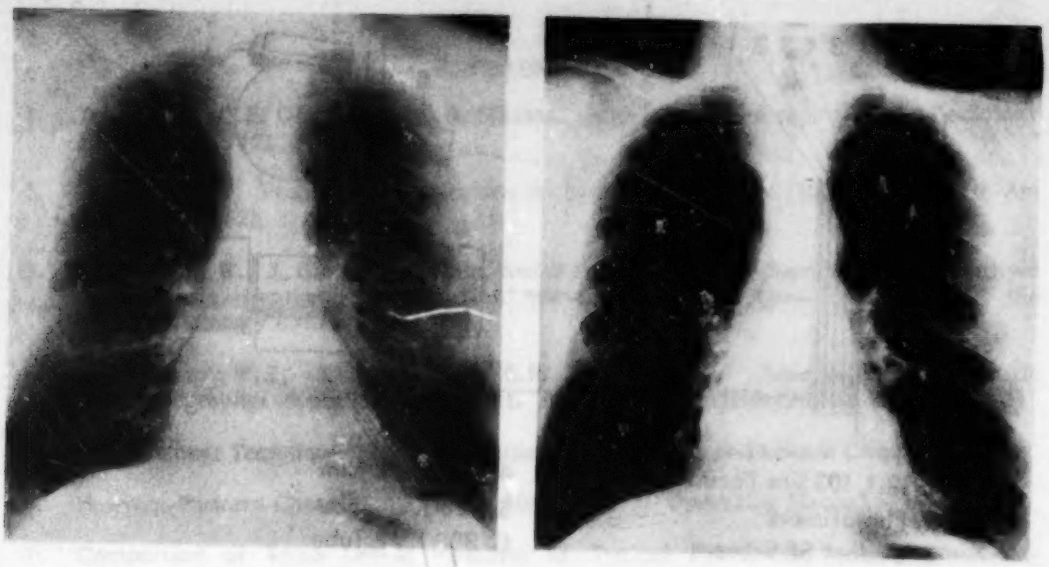
Figure 1. Diagramatic respresentation of the dual voltage tube and film grid and screen arrangement used in the production of 350 and 120 kVp chest films.



(A) 350 kVp

(B) 120 kVp

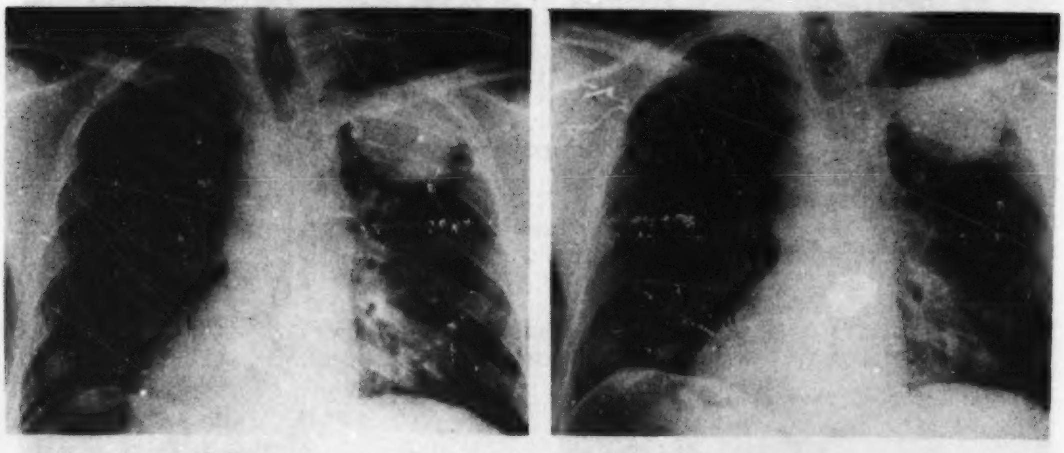
Figure 2. The calcification within the granuloma located in the left upper lobe in this patient is not as well seen with the 350 kV technique. However, with experience, calcification can be appreciated in most cases without additional films for clarification.



(A) 350 kVp

(B) 120 kVp

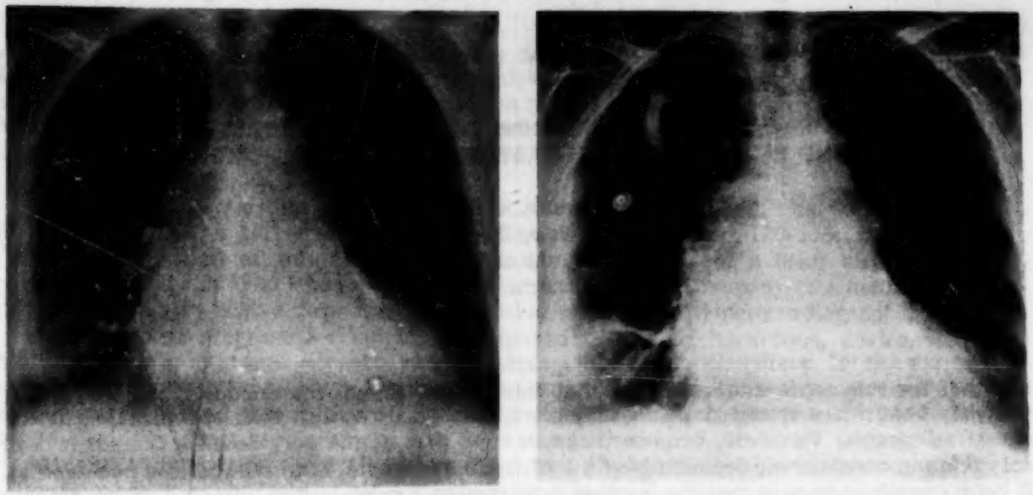
Figure 3. The 350 kVp technique allows more uniform distribution of pulmonary densities, as exemplied in this patient with chronic obstructive pulmonary disease. Bullous disease is over-penetrated with the 120 kVp technique, requiring transillumination.



(A) 350 kVp

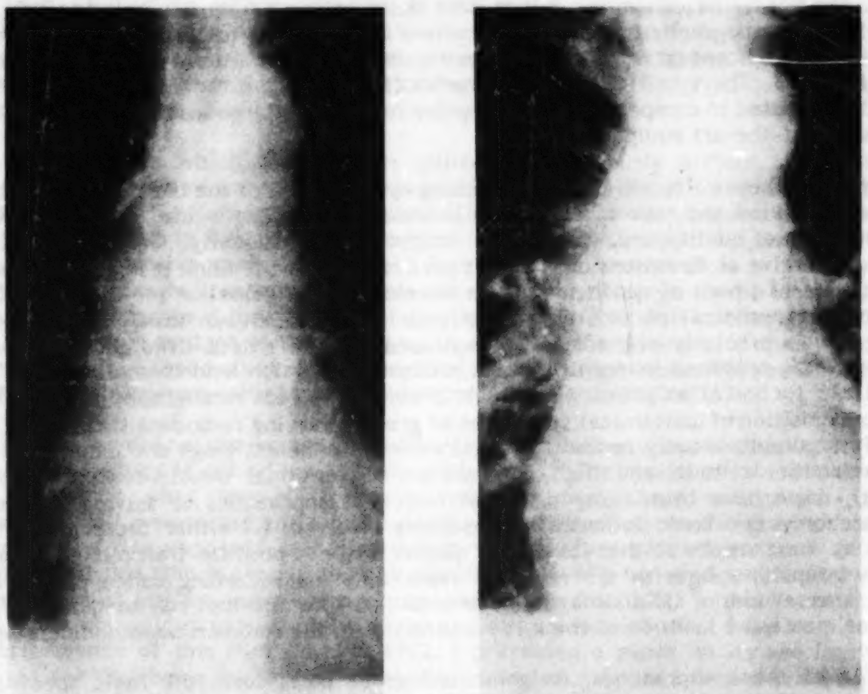
(B) 120 kVp

Figure 4. In this patient with metastatic hypernephroma, one notes better definition of many of the metastatic lesions with the 350 kVp technique. This is especially obvious in the pulmonary convexity (↑), near the left hilum (↑↑), and underlying the left clavicle (↑↑↑).



(A) 350 kVp (B) 120 kVp

Figure 5. Note the better definition of the prosthetic cardiac valve with the 350 kVp technique. There is poorer contrast definition of the metal snap in the lung periphery. This disparity is related to depth resolution.



(A) 350 kVp (B) 120 kVp

Figure 6. The 350 kVp technique allows better definition of the mediastinum and visualization of the dorsal spine, due to better depth resolution.

ALTERNATE APPROACHES TO EXTENDING LATITUDE, INCREASING DEFINITION AND ENHANCING CONTRAST IN CHEST RADIOGRAPHY

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The pursuit of quality in chest radiography is a challenging one and brings all of us together for this conference. I doubt that there is any other area of radiographic endeavor that has been more mired down in subjectivity and controversy than optimization of the chest radiograph. However, we must keep in mind that at the very best "optimum" is both relative and conditional, depending on a multitude of factors, both clinical and economical. It hardly needs to be said that the level of information needed by the clinician varies greatly and that we should have the flexibility to provide an optimum technique at an appropriate dose according to this need. Also, the patient's physical condition and habitus have a great bearing on what the optimum technical factors should be - obviously, the ideal film-screen-grid-focal spot combination will not be the same for the morbidly obese as it will be for those of normal habitus. Even our routine "survey" technique, if implemented in a dedicated automatic system, can only be "optimum" for the average patient. This, of course, can hardly be otherwise, but it does describe something of the complexity of our task. Bedside radiography is another area where it is difficult to optimize, unless we define "optimization" in context with clinical need and patient condition. Economic factors also weigh heavily. The small clinic or doctor's office with limited financial resources can hardly be expected to compete in what is optimum with the large institution and its (ideally) latest state-of-the-art equipment.

In light of these difficulties in approaching optimization of the chest radiograph, I think it is useful to view the task as a situation in which we need to decide, first, where we want to go in terms of quality and, second, how to get there technically. Only the radiologist or clinician can give us directions on the first part of this task. I think it might even be useful if categories of levels of quality could be developed as a guideline for the technologist in selecting his technical parameters. The second aspect of our task, how to get there technically, is precisely where so much equivocation now exists. The equivocation exists because of the problems inherent in chest radiography, which lend themselves to a number of separate technical solutions. The basic problem in chest radiography is caused by the close juxtaposition of anatomical structures of greatly varying radiodensities, which make it difficult to simultaneously record the chest's contents (heart, lungs and mediastinum) with both extended latitude and high contrast in conventional double-coated x-ray films. Although there have been many different technical approaches to solving this problem, there are only two basic technical philosophies involved: (1) either the primary beam is altered by some means so that its photon quality will not produce transmitted information of such intensity ranges in the remnant beam that the recording range of the imaging system is exceeded; or (2) an imaging system can be selected that has adequate latitude to record or match the latitude of the x-ray intensities in the remnant beam incident upon it.

Either of these approaches, or combinations of both, form the basic spectrum that covers most of our efforts in trying to achieve optimum chest quality. The first approach includes: (1) altering the photon quality of the primary beam from the conventional 70-80 kVp range to intermediary voltages ranging from 100-150 kVp, (the most common) to an occasional 350 kVp; (2) hardening the primary beam through heavy filtration; and (3) altering the spatial distribution of transmitted photons by the construction of complex tunnel and wedge filters. The second approach includes the testing of multiple screen-film-grid combinations, in order to select those combinations producing the best quality at reasonable or lowest dose. Workers using both concepts have achieved excellent results, with the optimum quality being reached when the transmitted subject range equals the recorded one. Please note again, this definition of the optimum can be reached either sensitometrically or radiologically. Providing certain basic parameters are met, there are certainly many different valid techniques that will take us to the "optimum" quality that we are pursuing.

The first technique that I will describe utilizes a single-coated, blue-sensitive x-ray film, Kodak S.B. 54., a 12:1 grid and intermediary kilovoltages from 120-150. Before going on with a description of this technique, I would like to present a brief history of my experience with it. In 1960 I was working for Douglas Aircraft Company as a medical x-ray technologist in a very active dispensary. In addition to pre-employment radiographs, we had an extensive chest program for employees exposed to various hazardous, environmental conditions, such as grinders, burrers, sand blasters and chemical handlers. Many of these employees were very large--much larger, in fact, than I have encountered anywhere else--and they sorely taxed the ability of my equipment (200 mA, single phase, 124 kVp) to produce a quality chest radiograph at the then standard nongrid, 80-90 kV technique. My first attempt at improving these films was to go to a 124 kVp technique (which was as high as I could go) with a 12:1 grid. This, of course, increased my subject contrast so much that the latitude of my technique far exceeded the latitude of the standard double emulsion film available at the time, and I ended up having to make two exposures on many of these employees, one for the mediastinum and one for lung parenchyma at a lower kV, but I was at least getting the depth resolution I needed through the mediastinum. About this time, I read an article, written by a Dr. Gerhart Schwarz, on extended range radiography utilizing a low-gamma single-coated x- ray film. Dr. Schwarz recommended this film for radiography of the extremities, skull, spine, pelvis, and ribs, as well as for various soft tissue studies; he claimed the film was superior to high kV techniques in extended latitude work because it reduced contrast without a loss of differential absorption. He also claimed that its density scale represented the optimum adaptation of radiography to the optic physiology of direct transparency viewing. I would like to quote his explanation:

> "The density range which the eye can utilize simultaneously without strain when viewing transparencies extends approximaately from 0.2 to 2.7 and represents a density scale of 2.5. Conventional double-coated films, however, possess a maximum density of from 3.2 to 4.5. When the subject range which is recorded on the film is spread over a density range of 0.2 to 3.2 or more, a portion of it is lost to the observer. In order to transmit all of the recorded subject range to the viewer, it is necessary to compress it into the useful density scale of 2.5. This can be achieved by limiting the silver content of the film, i.e., by reducing the thickness of the emulsion. The optimum is reached when the transmitted subject range equals the recorded one. This produces maximum utilization of the recorded subject range..."(1).

Although Dr. Schwarz did not recommend this film for chest radiography, because of its low contrast, I decided to try it out on my 124 kVp grid chests. The results were immediately successful, producing what seemed to me a near optimum chest radiograph. The mediastinum was well penetrated and almost always an excellent air tracheogram could be demonstrated. The ribs appeared as hollow tubes, and the lung parenchyma was clearly defined. The use of this film is clearly contraindicated in low kVp non-grid techniques. The scatter, low though it may be, ruins the performance of this film. Dr. Schwarz must have been referring to non-grid chests when he recommended against its use. I was so pleased with the performance of this film that, in 1962, I presented a paper on its use in chest radiography at the Oklahoma Society of X-ray Technologists. At that time there were no double-coated latitude films on the market and I predicted that, when such films were developed, they would be superior to the single-coated film for chest radiography. Time proved me to be wrong. I was working at my present position at Saint Francis Hospital when both Kodak and DuPont introduced their double-coated latitude films. This was a long-awaited event in my life, and I proceeded to switch our survey chest technique from the single-coated film to the new latitude film. There were two other reasons why I hoped for the sucess of the latitude films; one was processing (the price of using Kodak S.B. 54 was a 4.5-minute processing cycle which created a "hassle" in a predominately 90-second department), the second was cost. Since very little of this film was being made, its production costs were naturally higher, and it bothered me to have to pay more for a film that had less silver in it.

Our results with the latitude films were disappointing. Although the new film was much better than conventional higher contrast films (Kodak R.P., Cronex 4, Curix R.P. 1), it did not match the performance of the S.B. 54, particularly in the mediastinum (Fig. 1). Consequently we went back to using the single-coated film and continued to do so until 1972, when we planned for our first fully automated chest unit. We had to abandon S.B. 54, because Kodak would not supply it to us in the noninterleaved form necessary for use in any automated system. As we were not satisfied with any of the double-coated latitude films, we reluctantly said "goodbye" to the 120-150 kV chest technique and purchased a field emission generator coupled to a DuPont Daylight Chest Unit. This was used until January 1979, when we switched back to a more conventional technique utilizing a standard 3 phase generator, double-coated latitude film and a 10-inch focal film distance with a 10:1 grid focused at 72 inches.

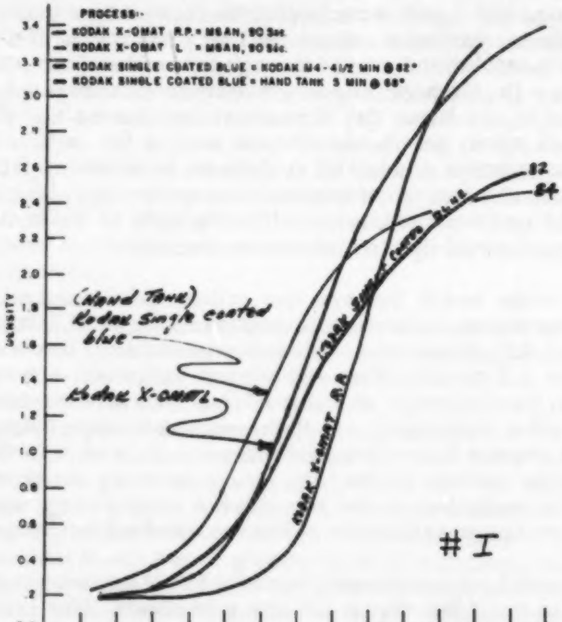


Figure 1

As a result of these technique changes over the past 8 years, we have been able to accumulate some excellent comparison films, using all three techniques on the same patients. We have some fairly strong opinions concerning each technique, which I will now share with you.

First, I would like to make a comparison of chest radiographs made at 350 kV with standard double-coated films and 120-150 kV with single-coated films. We were initially surprised at the close similarity between the two, especially considering a 200 kV difference between them. This similarity can best be explained by the unique sensitometry of the single-coated film. As the H and D curve on Figure 2 shows, the single-coated film has a very sensitive toe, developing good contrast in the region where little or no contrast is produced in normal double-coated films. It is precisely this area of the H and D curve where the low intensity (but spectrally harder) remnant radiation from the mediastinum is registered. At 120-150 kV, the latitude of the radiologic image exceeds the latitude of normal double-coated films and this low intensity radiation falls on point A (Fig. 3) on the conventional double-coated film, a region of no or very low contrast. However, the same intensity falling on point A of the single-coated film is within a range where, although low, very important contrasts are developed. This means that from a standpoint of large area contrast, the low contrast single emulsion film is better than the high contrast double-coated film! To emphasize this point, I would like to quote from a conference work by Dr. Robert Wagner and Mr. Earl W. Denney:

> "In general, however, it may be necessary to consider a wide range of exposure levels for comparing these two systems (high contrast and low contrast film). Subtle differences in the low density or "toe" region of a screen-film characteristic may give rise to large area contrast differences between two systems that would change their relative ranking from what it might be if the mid-exposure range were considered alone. A comparision will then depend critically on the technique and particular task in question..."(2)

A further examination of the sensitometric characteristics of the single-coated film shows a shouldering off of the curve at a maximum density of 2.5, which is also very desirable in chest radiography to prevent overexposure of the lung parenchyma. In the instance of the 350 kV technique, the latitudes of the chest x-ray images decrease to the point where they all fall within the high contrast area of normal double-coated x-ray film and produce good large area contrast.

It is thus seen that very similar results in terms of large area contrast and overall quality can be achieved by two very separate ways; one by radiologic means (going to 350 kV and compressing the latitudes of the chest images), and the other by sensitometrically selecting a film that has a much wider recording range.

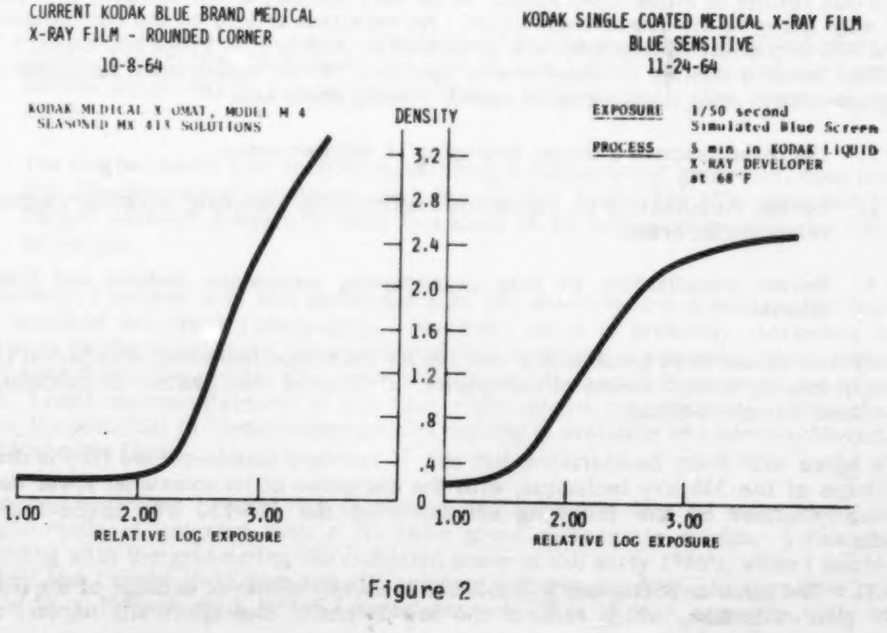
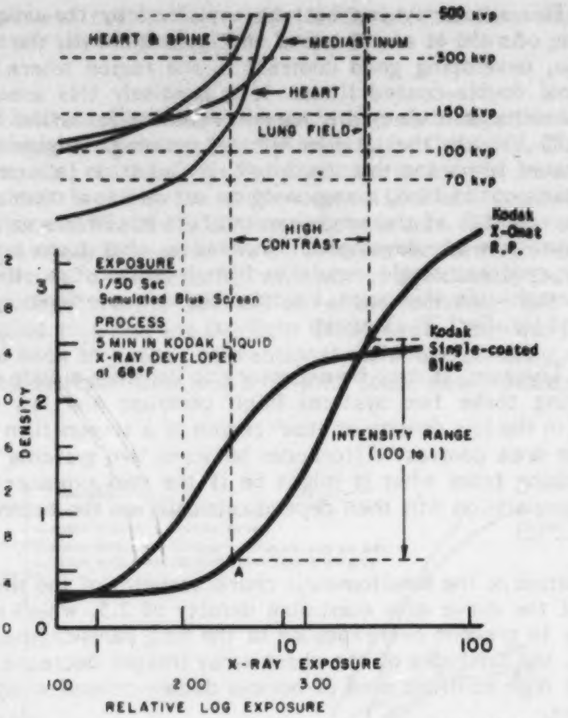


Figure 2



Compliments of
Dr. Charbonnier,
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Physical Aspects of
Diagnostic Radiology
Michel Ter Pogossian

Figure 3

The net results in either case appear to be very similar, in our opinion, with perhaps a slight edge going to the single-coated film. An excellent article on the comparison of 350 kV and 120 kVp chest radiography was published by Anthony V. Proto and Edward T. Lane (3). They listed a number of areas where they felt 350 kV radiography surpassed 120 kVp radiography (both using double-coated films). Among these are:

1. **Decreased exposure times, averaging 11 milliseconds.**

2. **Better visualization of mediastinal reflections, including infradiaphragmatic and retrocardiac areas.**

3. **Better visualization of lung parenchyma, pulmonary nodules and interstitial disease.**

They also listed three areas where the 350 kV technique had some drawbacks: (1) slight mottle, primarily in soft tissues of the thorax, (2) reduced visualization of calcium, and (3) generalized loss of rib detail.

We agree with these comparisons, but only if standard double-coated film is used. The advantages of the 350 kVp technique, with the exception of its somewhat lower dose rate, are mostly offset by the following attributes of the 120-150 kVp single-coated film technique:

1. **The same or better depth resolution is usually achieved because of the unique toe of this film, which records the low intensity (but spectrally harder) remnant**

radiation coming from the more opaque areas of the chest such as the mediastinum. In fact, since it is a well known fact that there is much greater latitude of the radiologic image at 120-150 kVp than at 350 kVp, there might possibly be more useful information recorded here than at higher kVp's (Fig. 3).

2. Better detail visibility is maintained in most chest images because of the intrinsically higher definition produced by the smaller focal spot and higher resolution screens sometimes used with this technique. Also, the film itself, having only one emulsion, probably contributes to the overall sharpness of this system. This is a subjective evaluation, and we are aware of several arguments that would seemingly negate our claim. These arguments are:

   a. The smaller focal spot is of no advantage over the relatively large (4.8 mm) focal spot of the field emission tube because the anode of this tube is conical in shape and produces a circular source, which has the resolving power of conventional thermionic tubes one half its size (4).

   b. There is "very little difference between the M.T.F.'s of detail and high speed screen systems when used with a 2-mm focal spot at 10 percent magnification" (5).

   c. The use of latitude films to extend radiographic latitude at 150 kVp techniques "sacrifices lung field detail to gain visibility in the mediastinum" (4).

Seemingly, then, we are obtaining results in everyday practice that theory predicts we should not. I think it is the subtle interplay between radiation contrast, film-screen contrast and the many factors which affect definition. According to Seeman, "an unsharp image boundary may be improved by increasing the contrast" (6). This we have done on our technique by dropping 200 kVp. "Conversely an invisible unsharp boundary may be made visible if it can be made sharper" (6). We have done this by going to a smaller focal spot and higher resolution screens and film. Seeman goes on to say that there is some kind of reciprocal relation between contrast and definition, but that it is impossible to predict how much contrast is equal to a given unsharpness, and vice versa.

3. Rib visibility, which is mostly lost in the 350 kVp technique, is maintained in the single-coated film technique, but not to the extent of masking parenchymal lesions which may be lurking behind them. Most typically they appear as hollow tubes.

4. The single-coated film technique, utilizing a conventional generator, does not run into "backup time" problems that the 40 MA rated 350 kVp generator does. Larger patients frequently take in excess of 65 milliseconds with the 350 kVp technique.

In summary, I believe that this particular film has sensitometric qualities that make it uniquely valuable for chest radiography. However, as it is presently marketed, being available only in the interleaved form and requiring 4-minute processing, it will appeal to very few; even those like myself who would like to use it, cannot, because of the way it is packaged. I hope the manufacturer of this film might see the potential value of his product and unlock its potential in chest radiography by making it available in a noninterleaved, 90-second processable form.

The second technique I would like to present utilizes a low-dose 3:1 ratio lead grid. A similar grid made of tungsten with a 5:1 ratio gives comparable results. I first began experimenting with the grid during the radiation scare of the early 1960's, when I wanted to see just how low I could go in dose and still produce a diagnostic film. At that time I could only get a 14 x 14 inch grid, which I taped over my 14 x 17 inch cassettes. My other exposure factors were a 0.3-mm focus, 30 mA, 124 kVp single phase generator and an average exposure time of 1/20 second. As I was limited to mAs flexibility in this technique, I often changed screens to control density, detail for small patients, par for average and high speed for larger patients. To my surprise, there was little gain in quality between the detail and the high speed screens. In fact, the high speed screens with the 3:1 grid gave better results than a detail screen with no grid at lower kV's and a larger 2.0-mm focal spot. I was surprised at the rather dramatic cleanup produced by such a low ratio grid. Dose rates, as close as I could measure them with my pocket dosimeter, ranged from 1 to 4 mR. I was pleased with the quality of these films, but I could never equal the overall quality of my higher dose single-coated film technique using a 12:1 grid. In particular, I could not penetrate the mediastinum adequately. I view this loss in quality as a result of the lower dose. I would like to recommend this technique for use in situations such as bedside radiography, surgery and emergency applications where a compromise in radiographic quality might be acceptable. In the case of bedside radiography, this might be the optimum technique because, in general, few technologists can manage to stay within the tight principal ray grid alignment necessitated by higher ratio grids. The low-dose grid has very wide positional latitude and is very forgiving in this respect. A 100 to 124 kVp chest technique with the low-dose grid is much preferable to a 70-90 kVp non-grid technique. Also, dose rates are much lower to the operator and, in the case of the surgical theatre, to other personnel who cannot avoid being near the x-ray beam.

The third technique I would like to present is called the long distance defocused grid technique. This technique utilizes a 10-foot focal film distance, a 10:1 or 12:1 ratio grid nominally focused at 60-72 inches and a conventional three-phase generator. We are currently using DuPont Hi-Plus screens and Curix latitude film. Our generator is coupled to a Philips Pulmo Diagnost changer and all exposures are ion-timed; maximum mA is 600, kVp is 140. I first saw this technique being utilized by Mr. Robert Takaguchi, chief technologist at Cleveland Clinic Hospital. I remember asking him why he was not using an air gap and was not concerned about his grid being severely longitudinally defocused. He replied that he did not use the air gap because of magnification and with this configuration he was able to radiograph all but the very largest patients on his automated system. As for the grid being decentered, he said "it is to my advantage because it increases the transmission through the patient's mediastinum." I have taken several radiographs of grids of different focal lengths defocused to 10 feet. The 12:1 focused at 40 inches is rather severe, and cannot be used unless it is reciprocated.

We have not yet received a grid to fit the reciprocating mechanism of the Pulmo Diagnost and are therefore using a stationary 10:1, 103 line, 60-72 inch focused grid taped to the front of the unit. This grid produces a very innocuous pattern, showing slightly inceased transmission through the center and only slight symmetrical attenuation bilaterally. What is happening is that the grid is acting like a tunnel filter to the remnant radiation coming from the patient incident to the image receptor. I might add that this creates a radiation pattern somewhat similar to the one produced by the complex mediastinal compensating filter made by Wilkinson and Fraser (7). However, this approach subjects the patient to twice as much radiation as the standard air gap technique and probably three times as much as the Wilkinson-Fraser technique. Nevertheless, I believe this is one of the most important survey techniques, particularly for automated chest units where the film cannot be transversed for large patients. The air gap technique reintroduces the same magnification present at 6 inches and at least some workers "feel that the imposition of an air gap results in some loss of resolution" (5). From a standpoint of minimal dose and optimum quality, Wilkinson and Fraser's technique seems theoretically nearly ideal, but I suspect not many workers will have the patience or ability to put together their very meticulous 18-part compensation filter, much less fabricate their special brass cone. Even then, assuming this is the ultimate technique, they will not be able to examine their larger patients because of the 10 percent magnification introduced by this technique.

There is another very important argument for the long distance grid technique that I would like to mention, which is of consideraable economic importance, particularly in the case of automatic chest changers where the film cannot be transversed. The very low magnification of this technique produces a near ideal geometry where it is not necessary to transverse the film for large patients, saving the time and cost involved in sending the large patient to another room for a non-automatic procedure. But this is not the main economic bonus of this technique. A 14 x 17 inch film format is no longer necessary, because there is almost never any relevant information recorded on the bottom 3 inches of the film. A 14 x 14 inch format is more than adequate. We tested this theory on 100 patients and only missed one. In a large department like ours, doing an average of 20,000 survey chests, this could amount to an annual savings of $10,000. I am aware of only one manufacturer who now provides for a 14 x 14 inch format on their automatic chest system. We are currently urging Philips to provide us with the engineering information to convert our Pulmo Diagnost to the 14 x 14 inch format.

In summary, a long distance defocused grid technique provides a nearly ideal geometry, avoiding retakes on the large patient and permitting the use of a 14 x 14 inch film format, enabling approximately 20 percent savings in film cost.

From a radiographic standpoint, the defocused grid contributes to optimizing chest quality by increasing transmission through the mediastinum and slightly attenuating symmetrically the remnant radiation reaching the lung fields. This radiation pattern, coupled with a double-coated latitude film at 120-150 kVp, produces good contrast throughout the entire chest volume that is almost comparable to the single-coated and 350 kVp technique previously mentioned. The main drawback to this technique is a slightly increased dose over the air gap. Our current entrance dose, based on LiF dosimetry, is approximately 14 mR for a PA exposure of a patient measuring 24 cm, exposure time approximately 8 milliseconds. We strongly recommend that all manufacturers of automatic chest units make an optional 14 x 14 inch format available for customers interested in this technique.

The last chest technique I wish to discuss utilizes a monochromatic blue color film that is practically grain free and has an infinite number of characteristic curves, varying with the spectral compositon of the viewing light. These two factors, plus the well known fact that visual acuity is accentuated by color led me to wonder if this film might not be an asset in chest radiography. I also felt that the merits of color in chest radiography, if any, should be brought before this symposium. Color radiography is not new, having been introduced in this country by Agfa-Gevaert back in 1970, but has never received significant acceptance. I understand, however, that it achieved considerable acceptance in Europe, particularly for angiography. The proponents of this film succinctly point out the problems inherent in black and white emulsions, the same ones that plague us in chest radiography. The difficulty is in reconciling emulsion latitude (which is in essence the possibility of recording a greater or lesser amount of information contained in the x-ray image) and contrast (which governs the legibility of the recorded information). We are faced with the dilemma of losing part of the information contained in the x-ray image, so as to ensure maximum legibility, or recording all the information with a loss in legibility. This is exactly the dilemma faced in choosing between a 350 kVp technique or a 120 kVp technique with a latitude film. I would like to quote from an article by Dr. J. Bories:

> "One solution is to use high kilovoltages, giving an x-ray image with a short exposure range that can be recorded in full by the emulsion without straying into the saturation zone. IT MUST NOT BE FORGOTTEN HOWEVER, THAT THIS "COMPRESSION OF CONTRAST" IS EQUIVALENT TO A REDUCTION IN LEGIBILITY. Color coupling makes it possible to separate the two functions of the emulsion (recording of photons and visible transcription of the absorption): the latter is left to the silver halide, but the transcription function is left to the dye..."(8).

That is about all the time I have to devote to color sensitometry. I wish I had had more time to work with this film, but I think the few results I have here will give you some idea of its potential. Exposure factors were 140 kVp, 6-12 mAs, 12:1 grid and Agfa-Gevaert Rare Earth MR 200 and 400 screens. There seemed to be very little, if any, gain in going to detail screens with this relatively grain-free film. Ideal viewing of this film requires variation of the brightness as well as the spectrum.

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SELECTION OF EQUIPMENT FOR CHEST RADIOGRAPHY

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It has often been said that the primary reason for selecting any piece of x-ray equipment is the availability and quality of maintenance. If the machine doesn't work, you can't perform x-ray examinations. If you can't select the optimum piece of equipment because the manufacturer does not provide the proper availability and quality of maintenance in your area, you will be better off selecting less than optimum equipment that is maintained in a quality fashion.

The workloads in chest radiography are extremely large. About half the examinations done in the United States are radiographs of the chest. In general, the design of the radiology department has not taken that workload into proper consideration. It is easy to justify and is cost effective to have radiographic rooms absolutely dedicated to doing chest examinations. On the same basis it is easy to justify x-ray generators and pieces of x-ray apparatus that are absolutely dedicated to 50 percent of your work. The concept of departmental patient flow patterns is important. It is really not proper to object to dedicated chest units integral with the processor because it isn't possible to examine a patient on a stretcher or a patient with a wide chest requiring a crosswise orientation of the cassette. Those few patients should be sent somewhere else to get their chest x-rays; 99 percent of chest examinations can be done in the specialized room.

Chest x-ray examinations are what radiologists do most and they ought to be done in the proper sort of room with the proper sort of equipment.

Radiologists do a lot of repeat examinations or extended examinations such as tomography partly because the routine use of stereoscopic chest x-ray examinations is out of favor. Research indicates that the accuracy of diagnosis--the sensitivity of the examination--could be improved by 15 percent if radiographs were made with a stereoscopic tube film shift even if they were not viewed stereoscopically. That is, if they are simply viewed side-by-side, in one the lesion will have moved away from the structured noise background so that it is easier to detect. Stereoscopic radiographs give twice the dose except that it is much smaller than the dose required for tomography and other examinations in order to increase diagnostic certainty.

Most machines have satisfactory line voltage regulation so that kilovoltage of the tube is related, usually with some degree of closeness, to the meter indications. The resistance, and particularly conductance in the line, is generally not as important for conventional x-ray machines for chest radiography. Considerations of conductance are usually important with angiographic examinations with large exposures at rapid rates, but the quality of chest radiography can be improved by using a much more powerful generator than is frequently recommended, making considerations of power supply more significant.

Conventional wisdom suggests that for chest radiographs a low capacity, single-phase generator can be used and that three-phase is not necessary. There are reasons to believe that this is incorrect. Three-phase generators decrease the dose to the patient and produce the requested radiographic exposure with an overall shorter exposure interval, thus decreasing the motion unsharpness in the system. Many of the recommendations concerning chest radiography suggest short exposure times to decrease cardiac or vascular motion in the chest.

There is not much difference, in terms of the ripple, between three-phase 6-pulse and three-phase 12-pulse. The real advantage is the greater stability of 12-pulse generators. The traditional 6-pulse generator with six rectifiers with a primary Delta and a secondary "Y" configuration of the transformer, has a very serious disadvantage of floating "ground" because of the difference in phase of the voltage and the current in the three-phase circuit. Thus, "ground" varies above and below zero. A stable ground can be accomplished only by putting in another circuit with an inductance and a capacitance in order to damp these oscillations. Lack of this circuit led to the instablity of early three-phase generators and to their failure of acceptance, in the U.S. at least. When solid state rectifiers became available, all the problems of having vacuum tubes and their filament transformers in the high voltage oil tank were eliminated. This made possible a 12-rectifier design with a primary Delta winding and two secondary windings, one "Y" and one Delta, so that ground was definitely "zero" and the problem of a floating ground was eliminated. That's really the advantage of the 12-pulse generator. They are more expensive, but are probably worth the expense.

There is substantial confusion about x-ray generator power. Most manufacturers now specify performance of their generators in terms of kilowatt rating. The radiologist says, "I would like a 150 kV, 1000 mA," and then is devastated when he can't run it at 1000 mA at 150 kV. That's simply because the radiologist doesn't quite understand what that means. The BRH, our friends in Washington, have taken care of making the manufacturers toe the line in terms of accuracy of kV and kV increments, but the radiologist has to know what the kilowatt ratings mean because you can buy generators with the same kilowatt ratings but different combinations of kV and mA. Confusion will result if you consider only the maximum kilovoltage and maximum milliamperage of which the machine is capable. The kilowatt rating of a 3-phase machine is calculated by multiplying kV times mA and dividing by 1000. With single-phase, you have to include a 0.7 constant factor in order to get the kilowatt rating.

Conventionally, kilowatt specifications of generators are based on a rating at 100 kV, e.g., a 70 kW generator will produce 100 kV at 700 mA, an 80 kW generator will produce 100 kV at 800 mA, a 100 kW generator produces 100kV at 1000 mA, and a 150 kW generator produces 100 kV at 1500 mA. So a "150 kV/1000 mA" generator won't give you 150 kV at 1000 mA; it will give 150 kV at 667 mA. It is perfectly legitimate to describe this as a generator that has a maximum 150 kV and a maximum 1000 mA. The kilowatt ratings of these generators are all specified on the basis of the mA at the 100 kV level. Two generators that are rated at 150 kW can be different - for different, very specific purposes. The generators are rated at 150 kW because you can get 1500 mA at 100 kV. One generator has no 150 kV setting on it at all, but can be operated at 70 kV and 3000 mA. Another machine has 150 kV at 667 mA; at 75 kV you can get 2000 mA. The 70 kV/3000 mA machine is specifically designed to do cine radiography or spot film radiography of the output phosphor of an image intensifier, particularly with iodine contrast materials using low kV for fast exposures at high milliamperage. There is no setting at 150 kV because that would decrease the contrast of iodine contrast materials. Both machines are specified as 150 kW generators. You have to know what you want and what you are buying.

There are many factors one has to consider in terms of timing. One could go to automatic exposure timing/phototiming and forget much of this timing problem. The initiation delay or the interrogation timing of generators is usually most important for angiographic work. With a single-phase generator, because the timer must relate to the 60-cycle sine wave, there is a minimum of 8.3 ms after the button is pressed before the machine can possibly respond. With three-phase you can get times as low as 1 or 2 ms because of the overlapping phases in the three-phase generator. With a constant potential, presumably you can have an immediate exposure after you press the button or essentially no delay at all. This is not as important in chest x-ray examinations as in angiographic examinations, but it is not totally unimportant. The size of the heart changes remarkably as the patient holds his or her breath because you decrease the venous return to the thorax with deep inspiration; the more protracted this modified ValSalva maneuver, the smaller the heart becomes. Therefore, exposure initiation delay can be a factor if you're interested in heart size.

It is important to understand the effect of current on focal spot size. The focal spot size is given at a specific current. When current is increased, the focal spot has a tendency to "bloom" and enlarge. The lack of clinical perception of changes in focal spot size up to approximately 1.5 to 2 mm is indicative that in chest radiography this is probably less important a consideration than it might be in contact radiography. Obviously, in magnification radiography, focal spot size is a significant factor. With a focal fim distance of 2 meters, focal spot size is apparently substantially less of a problem.

There has been less acceptance of anatomical program generators in the U.S. than one would have anticipted. This is obviously one of the ways to decrease the number of repeated examinations - by allowing the decision of when the radiograph has had enough x-ray exposure to be made by a well-designed computer rather than through the random action of a technologist who should be taking care of the patient instead of adjusting the dials on the x-ray machine. One of the problems in acceptability has been that many of the phototimers and anatomical program generators offered by the manufacturers until recent years have been very poorly designed and have allowed operator changes in the density response of the phototimers rather than letting the computer make the proper adjustments.

One approach that may be utilized with anatomical program generators, if the shortest possible exposure time is a design parameter, is the use of the "falling load" principle. This is appropriate for most areas of the body other than the chest. For the chest, exposure times are all relatively short. "Falling load" means that you start the generator off essentially with its highest output and let it fall either smoothly or in steps along a curve that corresponds to the heat capacity of the x-ray tube. What happens if you don't do this and have an anatomical program generator? Well, then you have to select a compromise mA that will take care of the average patient so that you will always have enough mA for the biggest patient without exceeding the tube load. An average mA is selected, resulting in an average exposure tme in the average patient rather than the shortest possible time. With "falling load", you get the shortest possible exposure time for each patient. With chest x-rays this is less important, because the chest x-ray exposure is usually completed within the first 100th of a second or so anyhow with a reasonable generator. By that time the "falling load", if it is a step one, is still at the first step; even if it is analogue programmed, it has not moved down from the top mA that is selectable.

COST CONTROL POSSIBILITIES

A number of years ago some manufacturers began to bring out modular generators. The concept fits in well with the principle of designing x-ray machines to do specific examinations, and do them as well as possible. Nothing demonstrates this principle better than chest x-ray examinations. By simply purchasing the modules required, generator controls can be tailor-made for chest radiographs. Recently, one manufacturer has made available a multi-room, multiple console device with a single generator. That cuts down on the cost of the generator, and any one of these multi-room consoles can be made into anything desired by module selection. A disadvantage is that, if the generator fails, several examining rooms fail until it is repaired. Fortunately, x-ray generators don't fail very often; but they do on occasion.

SUBJECTIVE FILM REQUIREMENTS TRANSLATED TO QUANTITATIVE SPECIFICATIONS

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INTRODUCTION

Consider the following suggestions that have been made to describe acceptable pneumoconiosis films:

1. Pulmonary shadows visible out to the periphery,
2. Costopleural junctions clearly visible, and
3. Larger pulmonary vessels visible through the cardiac mass.

I was interested in finding out in a preliminary way what these suggestions might imply in terms of densities on the radiogram, and in terms of scatter suppression. Densities were measured at selected locations on seven films. The films were five candidate normal radiograms that have been circulated for purposes of choosing example chest radiograms in the I.L.O. rating system for the pneumoconioses as well as two additional original radiograms supplied by George Jacobson. The five candidate films were copies with a rather high base fog level. This was removed by subtraction and a normal fog level (D=0.17) substituted for purposes of analysis. The positions selected for the measurements are presented in Table 1. Of interest was the density in the region of the mediastinum because this is a good indicator of the amount of scatter appearing in the radiogram (providing no image of the mediastinum is present). Also, the cardiac mass density increment above the mediastinum provides an indication of the visibility of the large pulmonary vessels in this area. The maximum lung field density was also measured.

Table 1. Densities and ratings of pulmonary vessel visibility in cardiac mass (worst film set equal to 1).

FILM	MAX DENSITY (FOG= 0.17)	MEDIASTINUM ABOVE BASE	CARDIAC MASS		RATINGS	
			FOG= 0.17	ABOVE MEDIASTINUM	G.J.	E.P.M.
J-1	1.73	0.02	.26	0.07	3.5	4.0
J-2	1.90	0.03	.27	0.07	3.5	3.5
1	1.64	0.09	.28	0.02	1	1
2	1.90	0.03	.24	0.04	3	2
3	1.72	0.04	.27	0.06	4	4
4	1.80	0.03	.23	0.03	2	2
5	1.70	0.03	.32	0.12	4	4

The large pulmonary vessel visibility in the area of the cardiac mass was evaluated by myself and G. Jacobson. These ratings appear on a scale of 1 to 4, with ratings 3.5 or greater being interpreted as satisfying or being better than the vessel visibility requirement. A plot of these results appears in Figure 1, where the density increase above scatter (ΔD) is plotted against rating. Note that the results correlate fairly well and indicate that a ΔD of about 0.06 is required. Note also, that if the same results are plotted using the absolute density of the cardiac mass area, and not ΔD, the results show no identifiable correlation with the rating. This is illustrated in Figure 2. There is certainly an indication here that a simple density specification for the cardiac mass area is not sufficient.

From the results with these film-screen combinations (whatever they were) and the techniques that were used, excellent chest radiograms, to satisfy requirement (3) above, should have:

a. A cardiac mass density increase above the mediastinum density of no less than 0.06, but no greater than 0.13; and

b. At the same time, the density in the mediastinum should be no greater than 0.03 above the base fog.

It is also of interest to note that there appears to be a correlation between the cardiac mass density increase above the mediastinum and the visibility of the costopleural junctions (requirement 2). This is tabulated in Table 2 and the results plotted in Figure 3. Note that, if the conditions (a) and (b) above are satisfied, the costopleural junction visibility requirement is also satisfied.

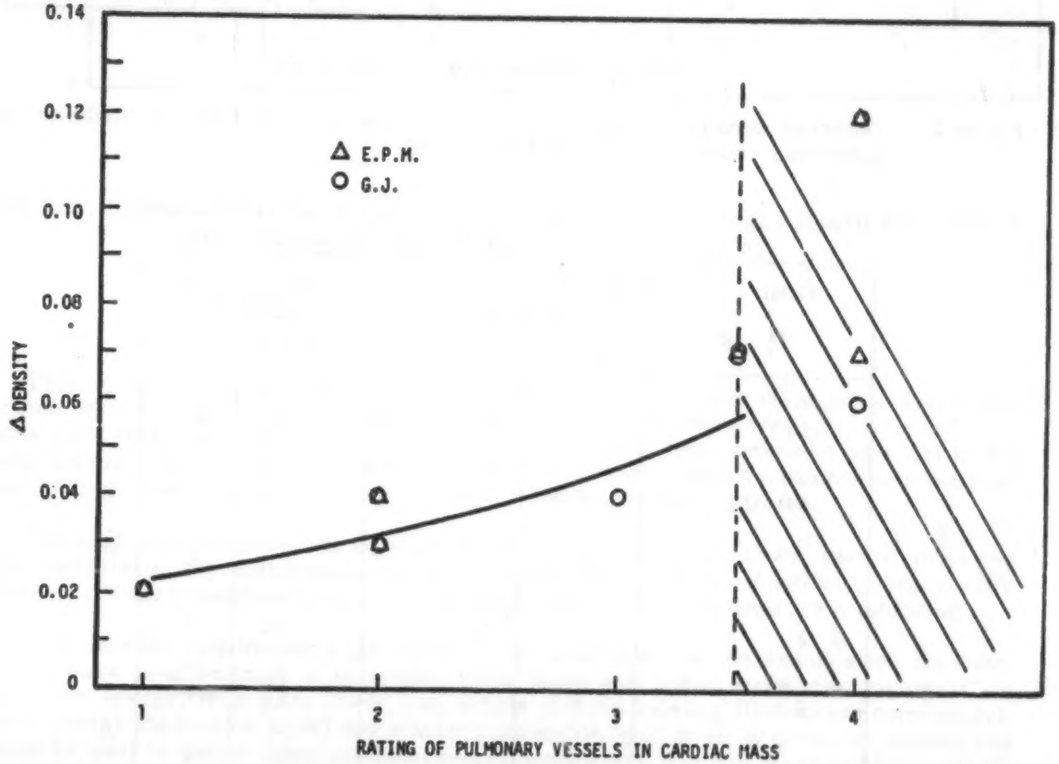

Figure 1. Observed ΔD between mediastinum and cardiac mass as a function of rating of visibility of pulmonary vessels in cardiac mass.

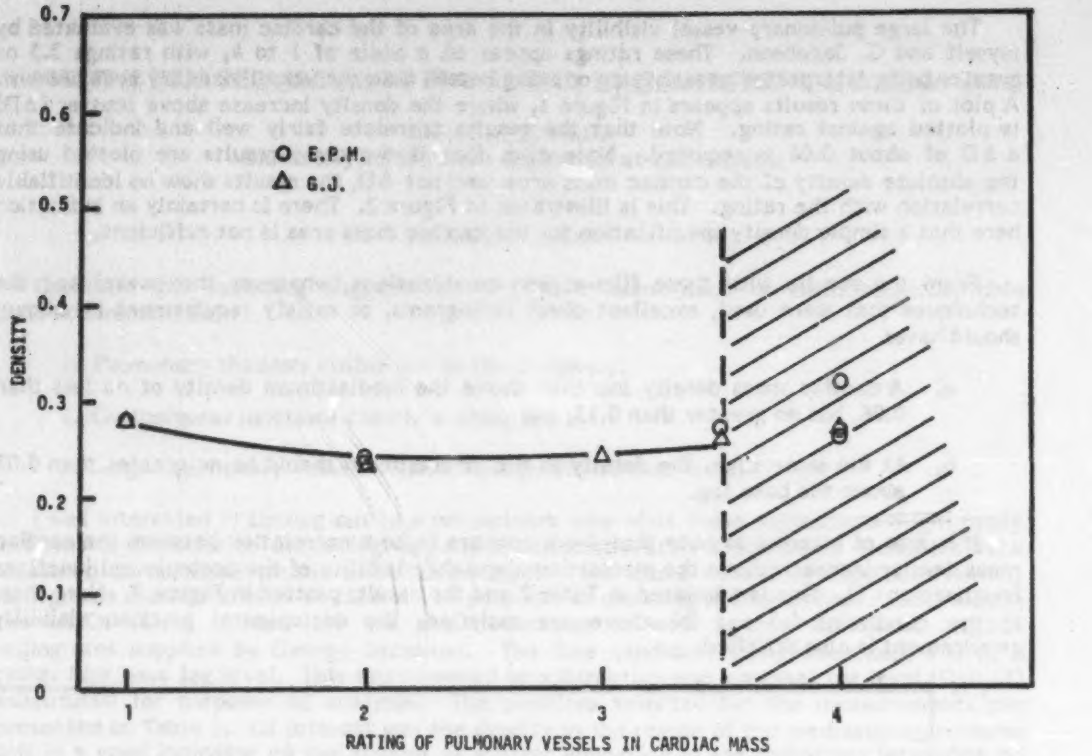

Figure 2. Observed density in cardiac mass as a function of rating of visibility of pulmonary vessels in cardiac mass.

Table 2. Ratings of costopleural margin visibility compared to pulmonary vessel visibility in cardiac mass.

FILM	PULMONARY VESSEL IN CARDIAC MASS		COSTOPLEURAL MARGIN	
	E.P.M.	G.J.	E.P.M.	G.J.
J-1	4.	3.5	3.5	4
J-2	3.5	3.5	4.	4
1	1	1	1	1
2	2	3	3.5	3.
3	4	4	4	4
4	2	2	2	2
5	4	4	4	4

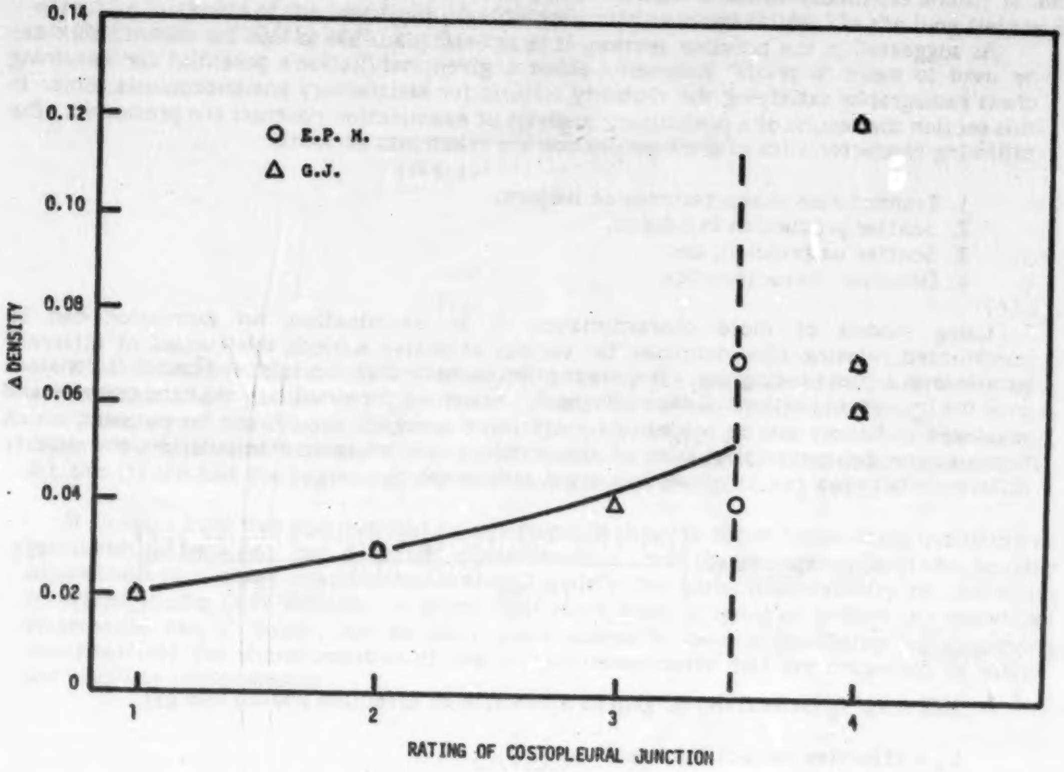

Figure 3. **Observed ΔD between mediastinum and cardiac mass as a function of rating of visibility of costopleural junctions.**

Finally, note that the maximum lung field densities for the three films which rate highest in the visibility of pulmonary vessels in the cardiac mass are: 1.75, 1.72 and 1.70. Note also, that three out of four of the lower rated films have maximum densities in the lung field of 1.9, 1.9 and 1.8. It would appear that in these cases the examination contrast was too high, due to insufficient beam energy or excessive film contrast.

The suggested requirements on visibility of pulmonary vessels in the cardiac mass and the visibility of the costopleural junctions appear quantifiable. Analysis of many more films, particularly positive films, is recommended before firm numbers are established.

The visibility requirements put rather close tolerances on the examination contrast. With a more detailed study of acceptable films taken with known screen-film combinations and beam energy, it is quite likely that simple criteria relating film-screen contrast (γ), beam energy (half-value layer) and scatter suppression (grid or air gap) can be devised and could be used to define those combinations of parameters that will meet requirements (1) through (3).

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PRELIMINARY MODELLING OF EXAMINATION CONTRAST

As suggested in the previous section, it is at least plausible to look for criteria that can be used to make "a priori" judgments about a given installation's potential for obtaining chest radiographs satisfying the visibility criteria for satisfactory pneumoconiosis films. In this section the results of a preliminary analysis of examination contrast are presented. The following characteristics of the examination are taken into account:

1. Transmission characteristics of subject,
2. Scatter production in subject,
3. Scatter suppression, and
4. Detector characteristics.

Using models of these characteristics of an examination, an expression can be constructed relating film densities for various effective subject thicknesses at different locations in a chest radiogram. The assumption is made that the scatter fluence is constant over the important portions of the radiograph. Assuming for simplicity that the primary and scattered radiation can be modelled by effective energies, results can be obtained which demonstrate the essential physics of the problem. After some manipulation, the density difference between two locations in a chest radiograph is:

$$D_1 - D_2 = \gamma_{eff} \log_{10} \left\{ \frac{e^{\mu(E)(L_2 - L_1)} \left[1 + 1.4e^{\mu(E)(L_1 - L_2)} / \Sigma(E)\alpha(E) \right]}{1 + 1.4e^{\mu(E)(L_2 - L_1)} / \Sigma(E)\alpha(E)} \right\} \qquad (1)$$

Where

$\Sigma(E) = T_P/T_S$ (selectivity of grid as a function of effective photon energy)

L_1 = effective subject thickness

L_2 = effective thickness which describes scattering from subject (assumed constant)

E = effective photon energy

$\alpha(E)$ = ratio of primary energy absorbed in detector per unit of entrance energy to scatter energy absorbed

γ_{eff} = effective slope of the characteristic curve between densities D_1 and D_2.

Apart from relatively large density differences that can be used to specify latitude limitations, local or point changes of density per unit of change in effective subject thickness is of interest because this describes the visibility of given local tissue-type targets. We can write:

$$\left| \frac{\partial D}{\partial L} \right| = \frac{\mu(E)\,\gamma}{\left[1 + 1.4e^{\mu(E)(L - L_S)} / \Sigma(E)\alpha(E) \right]} \qquad (2)$$

Now, if the ΔD between the cardiac mass and the lung field is held constant, equation (1) becomes:

$$\gamma_{eff} \log_{10} \left\{ \frac{e^{\mu(E)(L_C - L)} (1 + 1.4/\Sigma)}{1 + 1.4e^{\mu(E)(L_C - L_L)} / \Sigma} \right\} = K_1 \qquad (3)$$

Where $\alpha(E) = 1$, $L_s = L_l$, L_c is the effective thickness of the cardiac mass and L_l is the effective thickness of the lung field. Additionally, for constant $(\partial D/\partial L)$ in the lung field and another constant $(\partial D/\partial L)$ in the cardiac mass:

$$\frac{\mu(E)^{\gamma_c}}{\left[1+1.4e^{\mu(E)(L_c-L_l)}/\Sigma\right]} = K_2 \qquad (4)$$

$$\frac{\mu(E)\gamma_l}{\left[1+1.4/\Sigma\right]} = K_3 \qquad (5)$$

If all characteristic curves are assumed to be of the same shape, only differing by a generalized slope parameter γ_{eff}, then γ_l and γ_c become functions of γ_{eff}. In this case, the equations (3), (4), (5) have three unknowns (γ_{eff}, Σ and E) and can be solved. Other assumptions to truncate the series of simultaneous equations, of which equations (3), (4), and (5) are just the beginning, are of course possible and are being studied.

It appears from this analysis that it is reasonable to think about formulating quantitative relationships between the detector characteristics, the photon energy and the scatter suppression in a chest examination that will satisfy the subjective visibility requirements described in the first section. A great deal more work is required before any definitive statements can be made, but at least there seems to be the possibility of specifying quantitatively the characteristics of the system components that are necessary to satisfy the visibility requirements.

CAVEATS

This work is very preliminary. I have made some probably outrageous assumptions and the whole thing is in the midst of being redone more rigorously, so many details are left hanging or ignored in what I have presented. I am attracted to the conceptual approach and believe it is valid, but take reasonable warning that specific details are subject to revision.

ACKNOWLEDGEMENT

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DISCUSSION

DR. KELSEY: Will you tell us something about the radiation dose with the technique you use for the single emulsion film?

MR. MILLER: The dose is very similar to what we are now using for our 10-foot chest technique with a 10 : 1 grid, in the neighborhood of 20 mR for a 22-24 cm chest. Of course, that varies according to the screen-film combination being used. We are changing to one of the lanthanum oxybromide detail rare-earth screens that will cut that dose in half with no loss of resolution.

DR. GREGG: I hear two things. The first is that you do not need too short an exposure to eliminate motion and the second is that you do. I am wondering whether cardiac motion does not enter into the picture and disturb some of the clinical interpretations. Can the panel tell us about it?

DR. MOSELEY: I said you ought to reduce motion, so I will take that point of view. Actually, there is no question that in nearly all of radiography the greatest contributer to unsharpness is patient motion, either involuntary or voluntary. On the principle that you would like the highest possible resolution, you should do everything reasonable to reduce the exposure time and minimize motion unsharpness.

I could also argue the other side of the question if you would like--that is to say, most of the time we are dealing with lesions of such size that the resolution is not as much of a constraint as one would think. Most things we see in the chest are as big as your head and it does not really make much difference whether we have motion unsharpness or not.

DR. TUDDENHAM: I would take issue with you on the idea that patient motion is the greatest cause of unsharpness. I think it is the contour of the part, in most cases.

DR. MOSELEY: You mean the edge unsharpness? Yes; I will not argue with you.

DR. TUDDENHAM: Does any one want to take the opposite side of the coin since Dr. Moseley has now covered them both? Dr. Steiner, do you have any feelings about the importance of cardiac motion?

DR. STEINER: With a very short exposure time, I would expect cardiac motion to play a very minor role in producing visual unsharpness. For instance, in the studies we were doing, we were using 0.01 second exposures. I doubt very much if there would be any significant loss of definition of large anatomic structures.

DR. TUDDENHAM: I would guess that the answer to the question depends on what structure you are most interested in recording, and conspicuously absent in the data presented so far today has been any classification of the types of findings people look for and the frequency with which they look for them. Until we have that kind of data available we really will not know what it is we are after. I think in terms of big areas of pneumonia and big areas of atelectasis and shifts of the mediastinum while Dr. Milne looks at the microvasculature; and they require different kinds of imaging.

DR. FIGLEY: Robert Dinsmore from Boston has developed a scheme for displaying the amplitude of heart motion on a chest radiograph. In a normal heart, it is about 1 or 2 mm between systole and diastole at the left ventricular border and several millimeters at most in the aorta unless the patient has aortic incompetence, in which case it can be 5-6 mm. I saw some of these films and was surprised that the central blood vessels did not show more unsharpness when the heart was moving nearby with an amplitude of a couple of millimeters, so it strikes me that the need for a short exposure to image vessels sharply is probably overestimated.

DR. TUDDENHAM: I think you are now talking about large vessels, where the gradien across the border is steep so you can reduce it a little without impairing perception. In regard to the microvessels, you would see a much more profound change.

DR. MILNE: Actually, I have found that even if you take a relatively long exposure of the chest, with the patient doing good breath holding, you get surprisingly sharp images of vessels down to about 1 mm. I think the reason is that during systole the vessel is "pumped-out" consistently to the same volume and is held in that position for a fraction of a second, then "drops back" in diastole. The "dropping back" in diastole does not cause blurring because the edge you record is the one seen when the vessel is at maximum diameter. In the same way, if you let the heart beat several times during an exposure you get a very sharp cardiac border, not a blurred one; although in systole it has dropped back from its maximum size, it has already inscribed the larger diastolic image. I think this is also the reason why the University of California at San Francisco is managing to do CT scans of the heart with a 20-second scanner and getting what looks like a sharp image. The reason, of course, is that it is sharp only at the maximum diameter of the chamber--what is recorded is, in fact, a "pseudo-sharpness." If you take a long enough exposure, you never see the blurred edge. If you took an exposure which began and terminated during the time of systolic expansion of a vessel, you might get blurring; but I think that would have to be a very specific piece of timing and happens rarely. So, in fact, it seems that the visual impression of sharpness is not greatly affected by increasing the exposure.

Dr. Steiner, in comparing 350 and 120 kVp chest films, I wonder if you have really looked at the beam spectra of the two setups. Equivalent kV may be a more appropriate measure than kVp. I believe it was Mr. Miller who remarked that he was surprised there was not a greater difference between the 350 and 120 kVp chest films considering the 200 kV difference in technique. Of course, in terms of effective kV, the difference is a great deal smaller.

DR. STEINER: You are right, of course--the intensity is relatively small in the high kilovoltage range, as shown in Dyke's article (Radiology 117:159-164, October 1975). Most of the beam intensity is in the range below 200 kV.

DR. CHARBONNIER: Any tube using a thick target to produce bremsstrahlung x rays will produce a continuous energy spectrum with most x rays at low energies. It does not matter whether it is a field emission tube or a thermionic tube. It is true that there are very few individual photons at 350 keV in the 350 kVp field emission system. There are equallly few 120 keV photons in the x rays produced by a 120 kVp thermionic system. If one analyzes the two spectra, or considers half-value layers in copper, there are significant attenuation differences between the 350 kVp and 120 kVp beams, particularly when one takes into account the gradual hardening of the beam as it goes through thicker parts of the subject such as the mediastinum.

Absorption coefficients do not vary uniformly with voltage: the further up you go in kV, the smaller the variation in absorption coefficient. Thus, if you go from a 90 kVp to a 120 kVp system, there is a great deal of difference in absorption coefficients as a result of the 30 kVp difference. On the other hand, there is a relatively small difference between the absorption characteristics of 250 and 350 kV x rays. Still, when you compare 250 or 350 kV with 120 kV films, there is a significant difference.

DR. MUNTZ: I think we should not confine our discussion to kVp. I believe the system contrast is equally affected by the detector and the grid. To mention only kVp in comparing examinations is, frankly, in error. We must consider all three parts of the system and be very clear about what we say, or we will chase our tails around forever.

DR. ALLAN JOHNSON: I have a question for Dr. Charbonnier, or anyone using high kVp techniques. What sort of measurement has been made of filtration, beam homogeneity, or height of spectrum? Has anyone measured these beam spectra with a high resolution detector, a sodium iodine detector, match filters, or in any other way?

DR. CHARBONNIER: Field emission systems are generally characterized by their peak voltages, as required by BRH. These are 385 kVp for the stationary system and 240 kVp for the portable system. The high voltage applied to the x-ray tube consists of a train of very short pulses at a rate of 1,000 pulses/second. The pulse voltage waveform is not a perfect rectangle but has finite rise and fall times which affect the x-ray energy spectrum. The x-ray intensity spectrum has been calculated, based on measurements of tube voltage and current waveforms, and compared to the spectra for 90 and 150 kVp systems (see Ardran in Radiography XL: 277-284, 1974, for a comparison, and Dyke, Radiology 117: 159-164, 1975, for the 350 kVp spectrum alone). A broad spectrum is thought to be advantageous for PA chest radiography: the low energy portion helps maintain contrast in the lung field, whereas the high energy portion improves penetration through the mediastinum. The beam inherent filtration, approximately 5 mm of aluminum equivalent, yields a first HVL 50 percent greater than the minimum HEW requirements (17 and 11 mm versus 11 and 7 mm).

MR. FIELDS: I think a bit of advice is needed when specifying three-phase falling-load equipment. We found, in comparison studies of focal spot sizes for three-phase equipment starting with 1000 mA, that focal spot sizes exceeded the specifications by more than 50 to 75 percent. When we switched from a falling load to a fixed mA output (such as 100 mA), we found that specifications were met. We have had problems with some of our equipment where 0.3 mm focal spots are specified, but not for a falling load. With this equipment, the effective focal spot approached 1 mm with a falling load, but met the specifications at 100 mA fixed. In many of these instances where it is desired to take advantage of the small focal spot, we have been advising against using falling load.

DR. MOSELEY: I really do not disagree. There certainly is focal spot blooming with increased mA. Some designs minimize that problem, but obviously you will increase the size of the focal spot. In chest radiography, larger focal spots have a theoretical disadvantage. This is more theoretical than real, however, since focal spot unsharpness is not clinically appreciable in normally utilized systems.

DR. RAO: I am not quite clear about the reasons you gave for the use of three-phase equipment. After all, exposures as short as one-sixtieth of a second can be obtained with single-phase units, if desired. Secondly, with three-phase equipment, the exposure may be too short to insure good phototimer reproducibility. What, then, are the advantages of three-phase units?

DR. MOSELEY: Obviously, the shortest reproducible exposure time you can get with the single-phase machine is in the one-sixtieth of a second range. If you need a longer exposure, with a full-wave rectified single-phase apparatus, the next choice is one-thirtieth of a second, or two bursts. With a three-phase generator, the time for two bursts is much shorter.

Three-phase generation also has a spectrum that gives a higher electron voltage range, as we have been discussing here, so that the dose to the patient is decreased. The phototimer response of a properly designed three-phase system should be repetitive for exposures of 2 milliseconds or more; this is certainly short enough for chest radiography.

DR. SORENSON: Dr. Charbonnier, have you investigated the effectiveness of conventional grids when used at higher kilovoltages? Are they effective at all, or are they just a thin piece of lead in the way of the beam?

DR. CHARBONNIER: We have looked at the efficiency of grids and have made a number of comparisons based on grid parameters such as lines per inch, grid ratio and lead content per unit area, with both linear and cross-hatch grids. This is one of the reasons why I would see limited advantage in trying to go to voltages beyond the ones we have explored (240-400 kVp); the efficiency of grids decreases rapidly at higher voltages. At very low voltages you can consider the grid as opaque lead strips interspersed with relatively transparent spaces. As you increase the voltage of the primary beam, scatter tends to be more forward-directed and the energy of the scatter is much higher, resulting in incomplete absorption of scattered x rays in the grid. We have made measurements with phantoms using lead disks and a thickness of masonite simulating 3 or 4 inches of tissue. We find that with either a 12:1 linear grid or a 4:1 cross-hatch grid we reduce exposure behind this lead disk to about 10 percent of what it would be if we did not use grids, i.e., about 0.1 or at most 0.2 above fog level. This shows that grid efficiency is not perfect but is surprisingly good--better than I would have thought.

DR. HAUS: I would like to comment on Dr. Mosely's remarks about the effect of focal spot size in chest radiography. If the focal spot size is not important, if you can accept a large focal spot in chest radiography, then perhaps you can also accept a less sharp film/screen combination without markedly reducing total sharpness of chest radiography. For example, because of the large focal spot size of the Hewlett-Packard 350 kVp chest radiography unit, total sharpness is affected little by the choice of a "sharp" or "unsharp" film/screen system. However, when using a conventional unit with smaller focal spot size for chest radiography, you can improve total sharpness by also using a "sharper" film/screen combination. The combination of geometric motion and screen/film unsharpness is contributing to the total unsharpness in chest radiography. I do not believe that focal spot size is unimportant in terms of total system sharpness in chest radiography.

DR. MOSELY: There is no question, Dr. Haus, that we can demonstrate many things in the laboratory concerning the contributions of both film/screen combination and focal spot size to unsharpness, but my statement was that they are not clinically appreciable. At least, that is what I gained from Dr. Milne's presentation this morning. In many of his examples, unless you really get an extremely large focal spot and contact (nonmagnification) radiography, there is a great difference between clinical perception and the laboratory demonstration of the fact that you really do have unsharpness in the system.

DR. TUDDENHAM: I think the extension from focal spot requirements to film/screen requirements is a perfectly valid one. There are just some kinds of images for which technical unsharpness is not the dominant form of unsharpness and is not terribly important.

DR. HAUS: I would simply like to mention that there is another parameter--noise. When you speak of a fast film/screen system, noise enters into the interpretation. You could have two screens with the same noise characteristics but different appearances of sharpness or unsharpness of vessels.

DR. MUNTZ: May I ask a question about unsharpness? Why is it that we use whatever screen/film system we use; can we use much less sharp screens in chest radiography?

DR. TUDDENHAM: My guess is that nobody really knew what was required, so we use more than is really essential for the vast majority of things in the hope that we will never be caught short.

DR. MUNTZ: So the empirical evidence offered by the great wide world of radiologists voting with their feet to use whatever screen/film system they use is not necessarily good empirical evidence.

DR. TUDDENHAM: I think that is right.

DR. MILNE: Dr. Mosely, you quoted our experimental results correctly, but I do not think your interpretation is exactly right. For years I have said that I thought you would get better chest radiographs if you used 0.3 mm focal spots. Actually, the slide I showed does not disprove that. It shows a slightly different result. It shows that with a focal spot of about 0.5 or 0.6 mm, you do not see any deterioration as a result of an increase in focal spot size until the spot reaches about 2 mm; however, you definitely see an improvement when the focal spot is reduced from 0.5 to 0.3 mm. This is an interesting variation. I hope you remember that the vertical line on the observer analysis of the radiograph went from a 2 mm focal spot down to about 0.5 mm, then curved quite sharply to the right--which is toward the 100 percent level--meaning that every one of the observers noticed a definite improvement in the quality of the radiograph when going to a focal spot smaller than 0.5 mm. My statement was that between 0.5 and 2 mm you did not notice the slight degradation of film quality.

DR. TUDDENHAM: What happened when you changed the reading distance, Dr. Milne?

DR. MILNE: The reading distance? I did not try varying that. It was about 18 inches, but not standardized. I was trying to recreate conditions in which radiologists usually work, rather than create artificial conditions. Personally, I take my glasses off to look very closely and then I push back my chair, doing the "Tuddenham technique" of miniaturization.

DR. MOSELEY: With your glasses on? Did it improve the diagnosis or just the image quality?

DR. MILNE: We do not know that because it has not been specifically tested.

Dr. Tuddenham, you mentioned several times that you think the principal cause of unsharpness is vessel shape--the roundness of the vessels. Of course, one must agree with that. Kurt Rossman's work (Am J Roentgenol 85:366-371, 1961) showed that very nicely. Therefore, it is inconceivable to me that going to a higher kV, such as 350 kVp, could possibly result in sharper looking or better discerned vessels in the periphery of the lung. In fact, if you remove a lung, keep the vessels filled with blood, and radiograph it progressively from 5 to 10, 20, 60, 90 kVp, you progressively lose the peripheral vessels until at the highest kVp level they are not seen at all. They are "filtered," because there is no contrast left between them and the surrounding air.

DR. TUDDENHAM: I am not aware that anyone was saying the vessels would be better visualized at a higher kVp. I thought the object was to preserve them reasonably well and still do better in the mediastinum. Am I misquoting?

DR. CHARBONNIER: Two different situations must be considered. In the center of the lung field, which is relatively uncluttered and optimally exposed, the sharpness and contrast of blood vessels are obviously reduced at a higher voltage. However, this is not where the greatest visualization problems occur. What Dr. Proto reported in his comparative study of 350 kV and 120 kV films is that at 350 kV you could follow the blood vessels further toward the periphery and that the vessels were more easily, more continuously seen, even though their edges might be less sharp, because he was able to see them through overlapping ribs more easily. I think what we are saying here is that perhaps there is an enhancement of visibility of vessels which is not associated with greater contrast or edge sharpness but with the fact that you can see them more continuously than you would at low voltage. That is the way I interpret it.

DR. STEINER: Dr. Milne, in our study we found that the pulmonary vessels in the periphery were not as clearly seen and this was the comment we made. We are disturbed by it because we had quite a difference of opinion with Proto and Lane's paper (Am J Roentgenol 130:859, 1978). As Dr. Charbonnier said, it is a matter of semantics. We were looking for resolution and were not counting the number of vessels per area. We wanted to see interstitial densities well, and we felt we could not, compared to conventional technique. So we reported that interstitial structures were not seen as well with 350 kVp technique.

DR. TUDDENHAM: That may or may not be what people do in clinical practice. You are posing an arbitrary problem for them.

DR. RAO: In view of this discussion, what is the objection to the use of rare earth screens?

DR. CHARBONNIER: Rare earth screens, which represent a wide variety of materials and characteristics, generally offer increased speed, at the cost of increased noise, for a given resolution. The desired resolution and maximum acceptable noise level will thus determine the appropriate combination and will limit the speed which can be used. At the very high voltages where field emission systems operate (240-380 kVp), the speed advantage of rare earth screens is somewhat reduced while noise is more pronounced; hence, even greater care is required in the choice of film/screen combinations. Lanex regular screens with Ortho-G film, or Trimax 12 screens with Cronex 7, are examples of good combinations for high voltage applications.

MR. FIELDS: I would like to question Dr. Milne on how he measured focal spot size. Did he use a pinhole? If it was on a falling-load machine, it most likely was about 0.7 to .8 mm. We have measured hundreds of focal spots; they all come out like this when you run on a fallingload rather than a fixed mA.

DR. MILNE: Yes, I believe we are quite expert at measuring focal spots. We have developed a special focal spot camera which allows us to measure rapidly but with great accuracy. In our work, the focal spot sizes quoted were derived by pinhole measurements; however, in brackets beside these figures we placed the "equivalent" size, which is the true functional size taking into account not only physical dimensions but also the intensity distribution of the spot. Every one of the focal spots used was therefore measured with considerable care.

EQUIPMENT LIMITATIONS AND FILTRATION

Melvin P. Siedband
University of Wisconsin

Equipment for chest radiography falls into several categories as a function of the kind of traffic encountered. In a small hospital, a simple chest board will do. A large hospital can amortize more elegant apparatus on a cost-per-patient examination basis, and perhaps a fully automatic chest film system would be appropriate. Since one quarter to one-third of all films taken in hospitals are chest films, it is obvious that a high-traffic-flow system can pay off if it is controlled. Chest radiography is not a particularly high-exposure technique but, if it is improperly done, it can be a high exposure technique. It is also possible with chest radiography, particularly at the higher voltages, to add simple beam filters and cut the exposure down to low levels; 10-15 mR per exposure is not an unreasonable level to strive for and still get good films.

Quality control procedures assume that the apparatus was both designed and installed properly, but you will frequently find misaligned tubes or the wrong grid or the system is not properly calibrated. There is no way the gear can work at low exposure levels.

Another point is that systems are designed out of habit: If an older system used a 1.0/2.0 mm tube then the new system should use the same tube. Systems ought to be designed. The tube, generator, film holder, grid, and other components should be chosen to meet the requirements of each particular application. At the Unviersity we have several kinds of chest systems. Figure 1 shows a chest board made by Grafax with a phototiming pickup in the board; there is a knob at one side that permits the change of cassette size and orientation. The overhead tube can be used with this type of chest board. Figure 2 shows an automatic system. This is the Picker system which is coupled into the flm processor. There is a servo system so that, as the patient is positioned, the tube and collimator will track. The collimator can set field size; you do not always have to shoot the full dimension of the film. You might want to use the chest system for taking pictures of other body parts. In many of these systems it is of benefit to have a collimator that allows you to set a smaller field, especially for pediatric work.

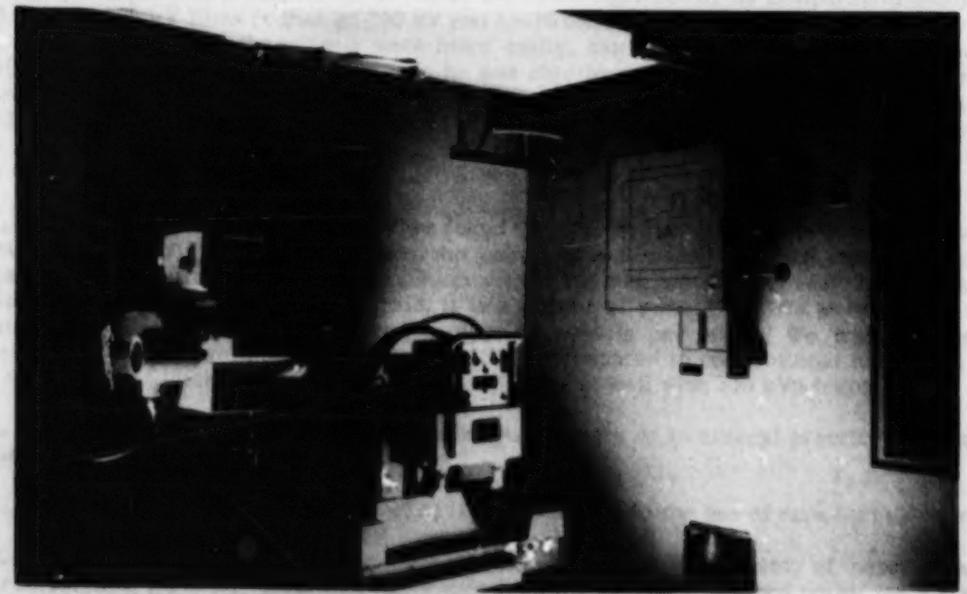

Figure 1. Grafax chestboard system.

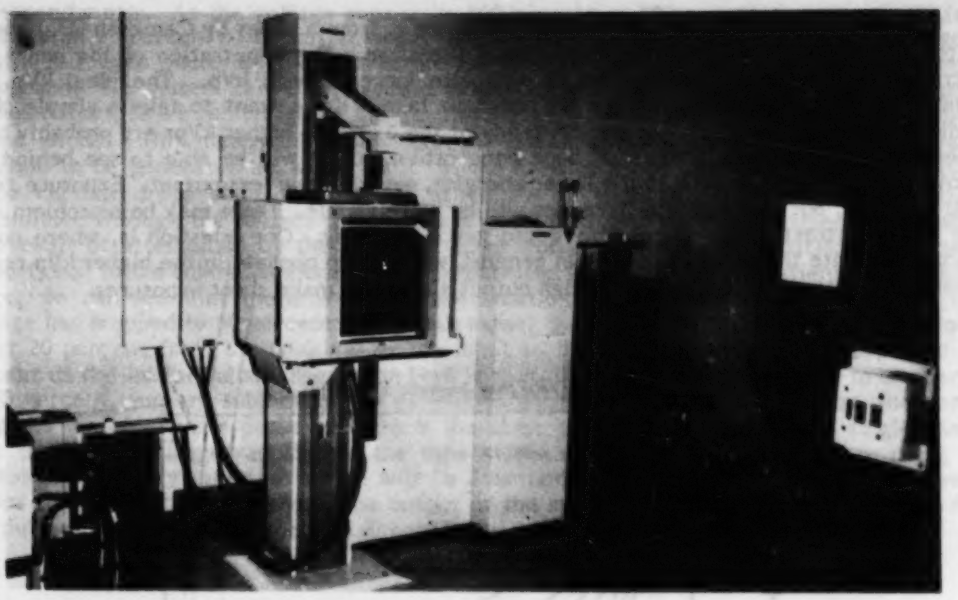

Figure 2. Picker automatic chest system.

Figure 3 is the Philips system; here it is also directly coupled to the processor which feeds through the wall to the film inspection area. This setup uses a mask system for a 14 x 17 or 11 x 14 field and will not be used for anything other than chests.

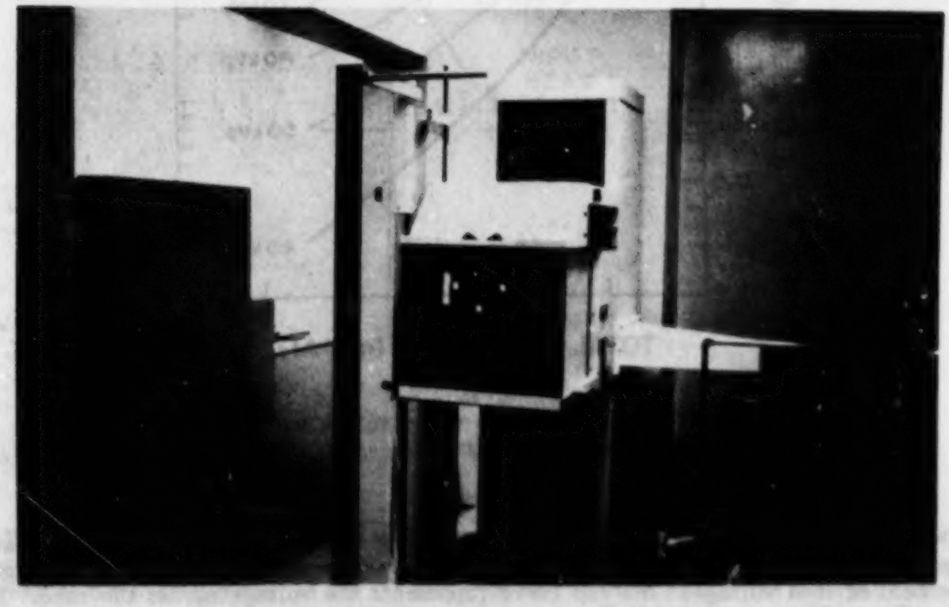

Figure 3. Philips automatic chest system.

We know from the output curve shown in Figure 4, from a paper by Cameron (1), how the output of a system will vary with the beam filtration. The penetration of the beam will increase with kVp; the system output will also increase with kVp. The ideal kVp is a function of the type of radiograph we want to take. If we want to take a simple chest radiograph to show geometry or show patency of the lung, the higher kVps are probably best. There will be a certain amount of bone penetration, so we will be able to see behind the bony structures. If we work at the lower energies, we will have less output. Exposure times will increase, but soft tissue differences will be more visible. There may be a problem with the bony contrasts that conceivably could get in the way. One question is, where do we want to operate the chest systems? In general, we want to operate in the higher kVp region where the system will have a fairly high output so we can make short exposures.

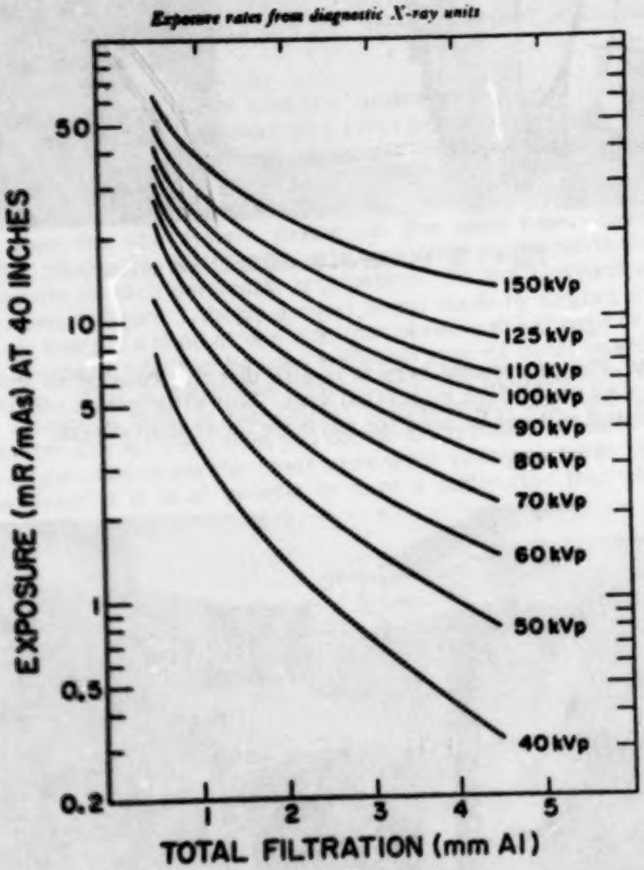

Figure 4. Output of an x-ray unit for various beam filtrations.

The output of the tube falls rapidly with decreasing anode voltage. Figure 5 illustrates this point. Assume a condenser discharge system starting at some voltage and dropping to 50 percent of that voltage. The top curve represents the voltage across the tube. The output due to Bremsstrahlung follows some constant times voltage squared and looks like the second curve. As the voltage goes from 100 percent to 50 percent, the output will fall to less than 25 percent of its initial value. If you are operating the tube at higher voltages, it will fall even more as the characteristic radiation output is diminished. In chest radiography the system is well above the Child's Law or Langmuir areas of space-charge limited output, so we cannot ignore that portion of the output due to the low voltage applied to the anode. This means that anode current is almost constant and low energy x rays continue to be generated at, say, kVp/2. In the case of a conventional system at 80 kVp, anode current drops about 50 percent at 40 kV dc instantaneous. The total energy transmitted through the patient to the film is going to follow the energy cubed relationship as a result of attenuation by the body. In other words, the body's attenuation is inversely proportional to the cube of the energy. If we examine the output of the patient, in the third curve, we can see that there is almost no radiation reaching the film by the time the voltage has dropped to 50 percent of its peak value. Indeed, if the voltage had only dropped about 20 percent from the peak value, the input to the patient would have dropped to 50 percent of the initial value. The lesson here is that, if you allow the capacitor to discharge to 50 percent, you are subjecting the patient to unnecessary radiation. But an additional lesson is that for single-phase equipment the same thing results, because in single-phase gear the lower voltages applied to the tube at the early part of the curve are going to produce low energy radiation which will be absorbed by the patient and produce no real effect on the film. Furthermore, the output of the system is less for single-phase than for the three-phase or constant-potential machines.

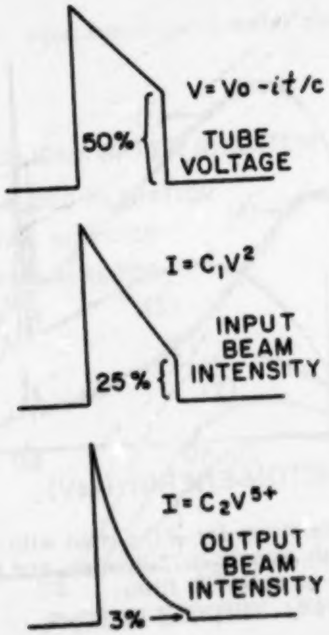

Figure 5. Illustration of x-ray tube output as a function of anode voltage.

The curve in Figure 6 illustrates the results obtained with a 2.5-mm aluminum filter in a diagnostic machine operating at 80 kVp constant potential or three phase. Curve #1 represents the overall output of the system, with no filtration except the inherent filtration of the tube. Curve #2 represents the output of the tube with the aluminum filter. Curve #3 represents that portion of the beam absorbed by the filter. This illustrates that the filter absorbs mostly low-energy radiation, certainly a much higher proportion of low-energy radiation than of higher-energy radiation. The intent is to reduce skin exposure and internal exposure to the patient, and not affect the output of the system to the film. The question is: How much filtration is enough? If 2.5 mm is good, maybe 5 mm is better. Figures 7 and 8 show the input and exit radiation to a water phantom as the aluminum filtration is increased. We notice that the output radiation form is almost the same. The output curves' energy distributions have the same shape but differ in amplitude. Notice that the energy distribution of the input from Curve #1 to Curve #2 changes as a result of two parameters: the proportion of energies represented at each energy level, as well as the integral of all the energies. What we would like to do is change the proportion of energies. We want to cut out the low-energy radiation but, if we start adding more than 2.5 mm of aluminum filtration, at 80 kVp, we notice an attenuation of the overall beam rather than an attenuation of the lower energies only. Since the relationship between Curves #2 and #3 at the output level is not much different than their integrals were at the input level, and since the difference between Curves #1 and #2 is far less at the output level than it was at the input level, one can see that about 2.5 mm of filtration is adequate. Additional filtration will attenuate the overall beam but will not greatly reduce the net skin exposure.

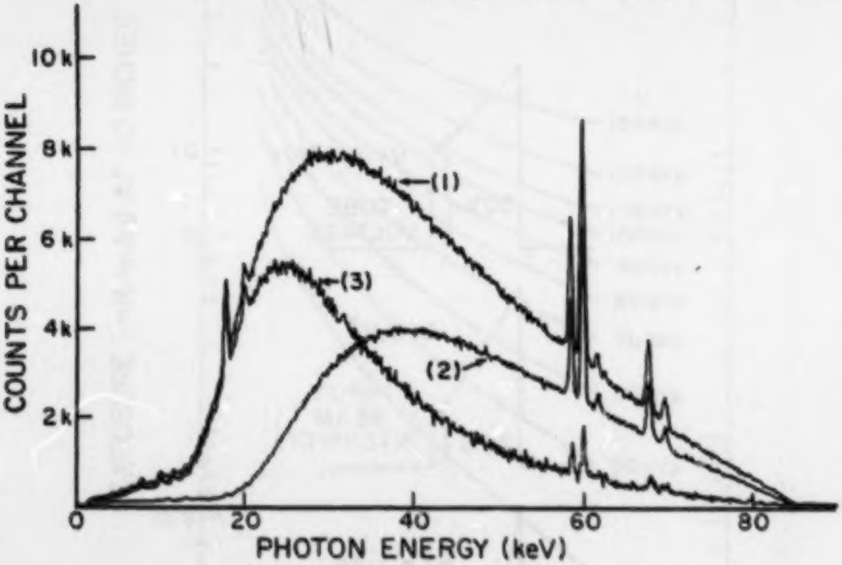

Figure 6. X-ray spectrum for (#1) output with no filtration, (#2) with aluminum filtration, and (#3) portion of beam absorbed by filter.

In brief, we found that optimum filtration is a function of kVp. If the Bureau of Radiological Health mandates 2.5 mm of added filtration at 80 kVp, it is not good at 60 kVp where the number should be less, and it is not good at 140 kVp where it should be quite a bit more. It is a simple matter to modify the gear for sliding in different filters. A Picker mammographic machine was modified to allow selection of a palladium filter or an aluminum filter for mammography, depending on whether we were using film-screen systems or Xerox. Such techniques in mammography can save about half the patient exposure.

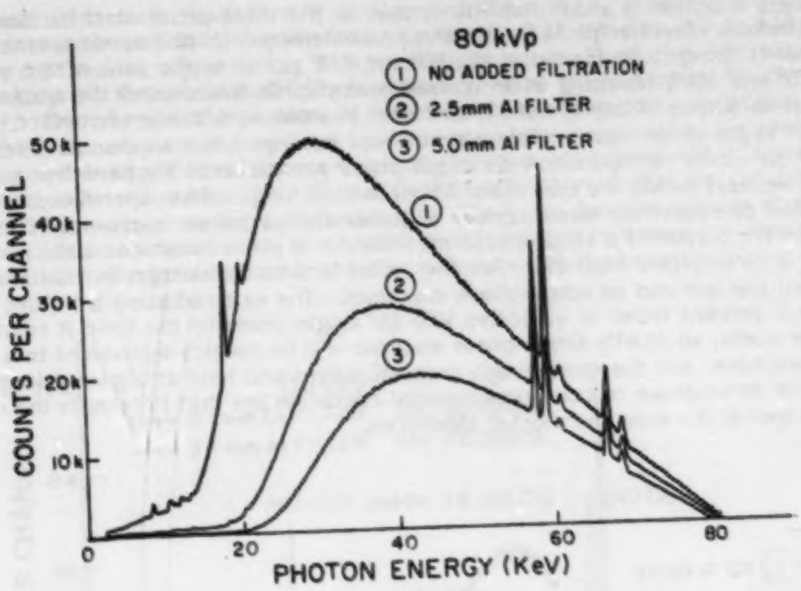

Figure 7. Input spectrum to water phantom.

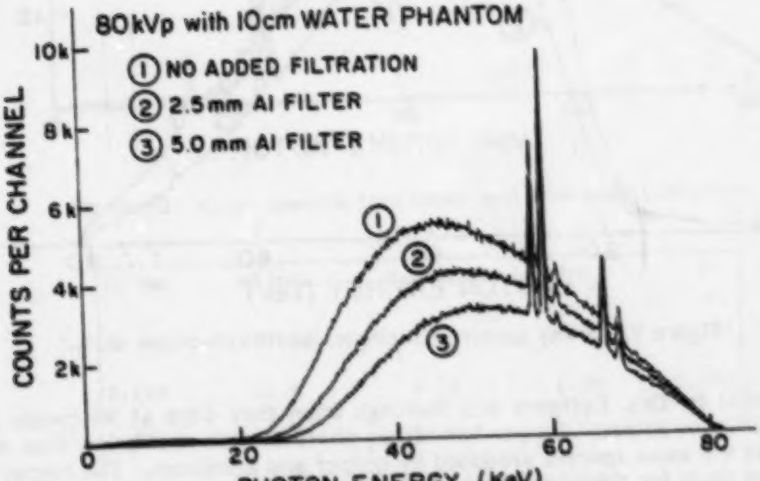

Figure 8. Output spectrum from water phantom.

Similar techniques exist in chest work. We know that the single-phase system is going to have more low-energy radiation than the three-phase system. Spectra, normalized for equal counts, are shown in Figure 9. We also know that two other factors make the single-phase gear a little less than optimum. First, it has more low-energy radiation output and the output is proportional to the integral of the square of the kVp. We know that the output of

the single-phase machine is about two-thirds that of the three-phase machine due to this energy distribution. Another point is that the pulsatile nature of the anode current means that the anode is going to be loaded at the 90° and 270° points at the peak of the wave and unloaded at 0° and 180°, resulting in an uneven heat distribution around the anode. Since the limitation on output of any system is how hard you can work the anode before it melts, we know that single-phase does not permit full heat loading of the anode. As a result, we must derate the tubes for operation on single-phase generators. We have less output in mR/mAs and we must derate the tube about 25 percent for single-phase operation over three-phase operation to obtain the same number of joules dissipation per exposure. This means that the effective output of a single-phase machine for a given focal spot size is less than half that for a three-phase machine. Another point is that the energy distribution is also skewed toward the low end on single-phase machines. The exit radiation from the patient will be about 5 percent lower in effective kVp for single phase by the time it reaches the film. In other words, an 80 kVp single-phase machine will be roughly equivalent to a 76-kVp three-phase machine, and the same rough proportionality will hold at higher voltages. The real benefits of three-phase or constant-potential operation are that it permits the use of a smaller focal spot in the tube and shorter exposures.

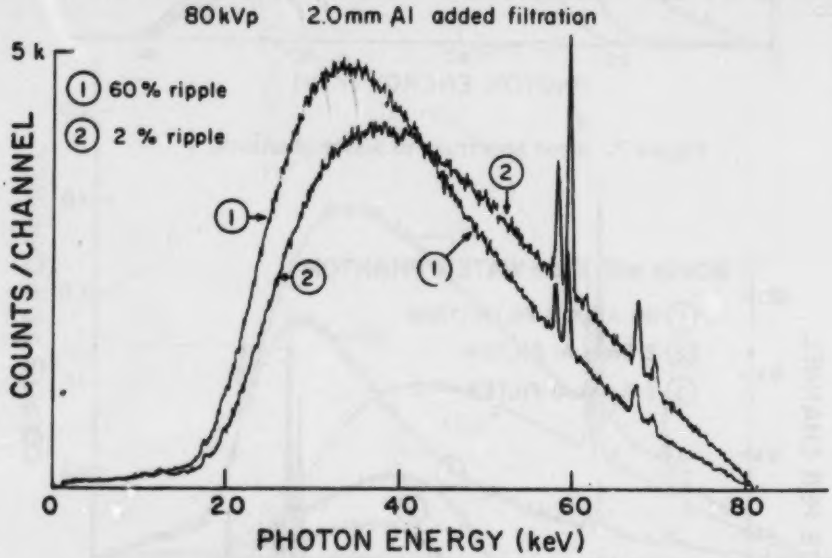

Figure 9. X-ray spectra for single- and three-phase units.

An experiment by Drs. Eastgate and Jennings when they were at Wisconsin compared copper and aluminum filters. Copper has higher photoelectric absorption than aluminum. Figure 10 shows the same spectra produced by copper and aluminum. The copper exposure took only 380 seconds for the same machine setting in our spectrum analyzer to give us an equal number of counts in our system. Because the copper was more efficient in letting the x rays through the filter--by that same ratio of 451 to 390--it is like getting a gift of more output from the system if you use copper instead of aluminum. This is particularly beneficial if you start going to higher filtrations. One of the factors examined was how much additional filtration would be best. A computer program was developed to find the relative contrast of a variable thickness object versus tube loading, exposure, and copper filter thickness for 120 kVp. Tabulated data are shown in Figure 11. We set an arbitrary criterion of additional tube loading in joules/exposure. The idea is that if the loading is above 40 percent of that level, you might have to go to a larger focal spot. The contrast level went from 0.227 to 0.221; in our phantom exposure this small change is not visible. Note that the skin exposure to the patient had dropped to roughly 40 percent of what it was before, and all it took was 0.25 mm of copper in the system. Chest radiographic systems should have some additional copper filtration—about 0.35 mm for 140 or 150 kVp is best. The first 0.25 mm of copper really does most of the work. To do this with aluminum would take roughly 40 times more thickness—about 10 mm of aluminum, which tends to be unwieldy. There is also the problem of inclusions in the aluminum, so copper foil is really best. If the same system is used for low-energy operation as well as for chest work, it is nice to have the copper on a flipper. One manufacturer has agreed and it is specified in our contract that he will add a little flipper in the collimator so that we can flip it into the beam for high-voltage radiography and out of the beam for low-voltage work. I suggest BRH give serious consideration to mandating that collimators used for chest radiography have some kind of solenoid to automatically add copper to the beam as soon as one operates over 110 kVp.

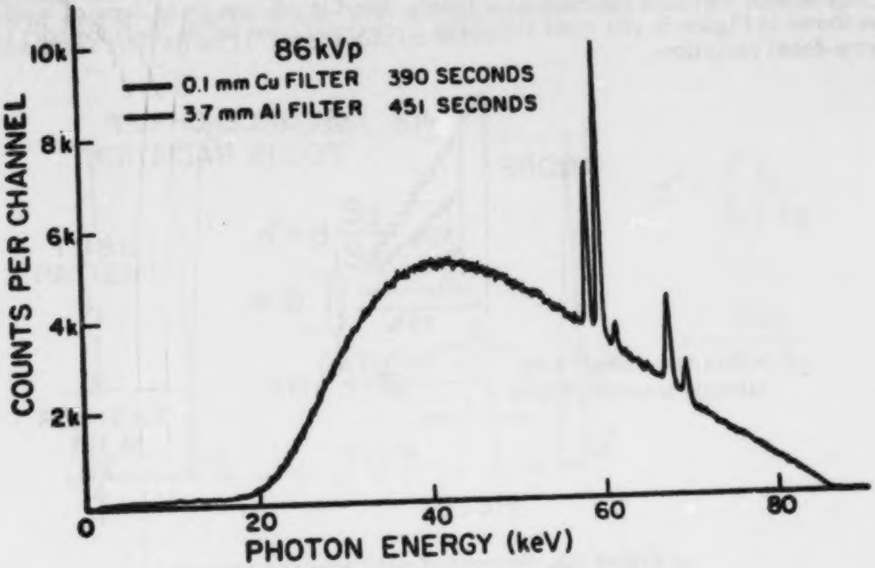

Figure 10. X-ray spectra for copper and aluminum filtration.

Cu mm	mR/mAs	Exp/Exp_o	W/W_o	C
0	27.0	1.00	1.00	0.227
0.125	11.4	0.51	1.20	0.224
0.25	7.80	0.40	1.40	0.221
0.375	5.94	0.36	1.63	0.218
0.50	4.83	0.33	1.88	0.217
0.625	4.05	0.32	2.14	0.215

Base: 120 kVp, 2.00 mmAl, CPG, Constant ABS. Energy
20 cm H_2O Phantom

Figure 11. Tube loading, exposure, and contrast levels for copper filtration.

Another problem concerns extra-focal radiation or off-focus radiation of x-ray tubes, as illustrated in Figure 12. Dr. Gapola Rao has measured the off-focus radiation of the tubes. Using less elegant methods, I have also made some measurements and obtained similar results which indicated that close to 30 percent of the radiation exiting an improperly coned tube will originate from somewhere other than the focal spot. It is almost as if most electron guns in x-ray tubes are similar to spotlight beams and there is a considerable amount of radiation off to the side. This means that the collimator has to do more than just define the field; it must have some type of proximal cone to reduce off-focus radiation by confining the source of the beam to the focal spot. If you use a simple cone on the chest system without some type of port cone, you will lose some of the contrast of the image. Contrast always appears in exposure formulas as an inverse squared term so that a slight improvement in contrast will produce a considerable benefit in patient exposure. A collimator is also supposed to predict where the beam will define the beam, and filter the beam; the mirror can also function as a filter. Even if you use some type of mask device, such as shown in Figure 3, you must still have a proximal cone inside the tube port to reduce the extra-focal radiation.

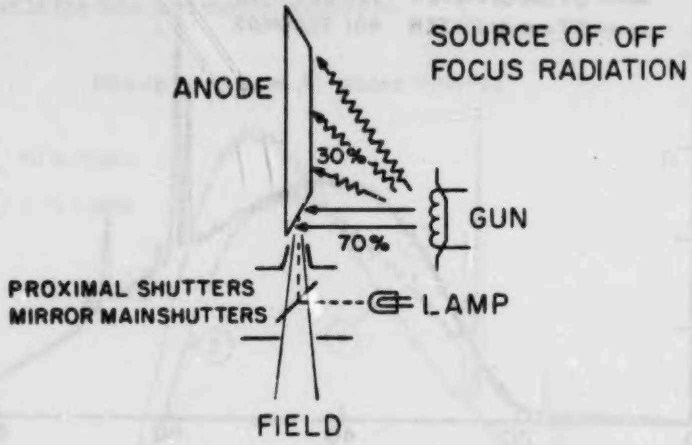

Figure 12. Sources of off-focus radiation.

There are several ways of reducing the effect of scattered radiation to the film. If you use an air gap, most of the scattered radiation will simply not reach the film. If the patient is in close contact with the film-screen combination, then some type of grid arrangement can be used. The grid works on the principle that the primary beam can go through the slots and the secondary scatter will strike the lead strips and not get through the system, but this demands that the tube be properly aligned to the axis of the slots. If you have a 12:1 grid and if the tube is off center by 1/12 of the focus-to-film distance, then the primary beam will be cut off. For example, if you have a 12:1 grid and you are 80 inches away, then if you are 1/12 of 80 inches off to one side (about 7 inches), you are going to have 100-percent cutoff; if you are 1/120 of 80 inches to the side (only 2/3 of an inch off), you will have a 10-percent cutoff of the primary beam. Thus, just an inch of misalignment can result in an increase of exposure of more than 10 percent, as well as a reduction of contrast because of the lower cleanup factor. I have seen several installations where tube alignment was off by far more than an inch; incredibly, one system was off by a full 5 inches! Another thing to keep in mind is that the real goal of an air gap technique is the reduction of scatter, which will follow the reduction in the solid angle of the received scatter measured at each point of the receptor plane. Obviously, the air gap technique works even better if you cone to a smaller field, because then the total amount of scattered radiation received by any point on the receptor is diminished. Air gap techniques, magnification techniques, and other related techniques work much better if the field is made smaller. The object is not magnification but reduction of solid angle, at the receptor.

If you do use air gap techniques, however, focal spot size becomes important. Figure 13 shows the effect of focal spot size. Assume that the focal spot size is projected through an infinitely small point within the patient and the plane of the film is of size d. If we then take that size, d_1, and project it back through the patient, that size corresponds to this smallest object within the patient, d_2. In plain language, this means that, if you are looking for small objects, the focal spot in the air gap application must be fairly small. Since a nominal 2-mm focal spot can be as much as 2.6 by 4.0 mm and be "legal" (NEMA or IEC), a 2-mm focal spot will often be too big. The tube should be chosen in terms of the angular coverage and in terms of the output required to do a particular job, and the smallest focus tube is best. The choice of focal spot size should also be a function of the exposure times we are willing to accept. The anode angle should be small, 11 or 12°; the system will be used at long throw distances to minimize magnification of the patient image on the film plane. Because the field coverage angle is small, we can have a higher rated tube with a smaller focal spot if we choose a tube with a smaller anode angle. A 17° tube is usually a poor choice chest system; an 11° tube would be better.

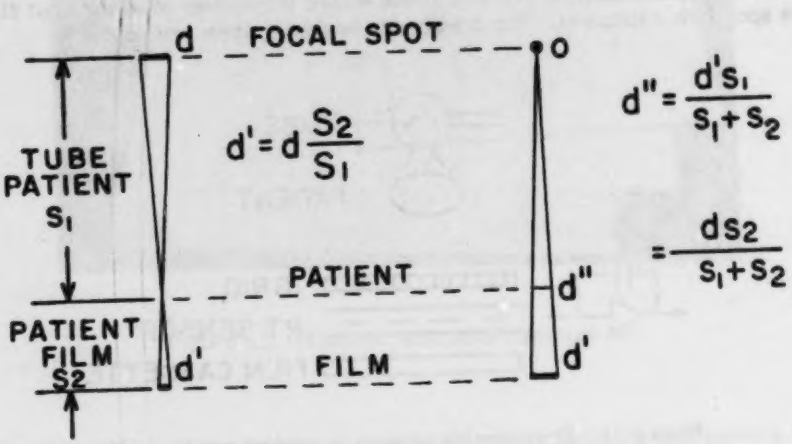

Figure 13. Geometric effect of focal spot size.

Anode heat capacity does not have to be very high because the heat generated during a single exposure is not very great and there is time for the tube to dissipate it. Heat capacity of the housing also does not have to be particularly high. If you take a great number of exposures per day, as ought to be the case with automatic equipment, you may want to put a fan on the housing in order to allow it to dissipate heat more rapidly. Almost all chest films are single-exposure techniques so that multiple exposure ratings of the tube do not apply. The maximum mA rating at low kV also does not apply. In general, systems should operate at high anode rotational speeds to permit the very short exposure times that should be used for chest radiography. Such systems do not require an intercooler; this device is a heat exchanger with a fan, similar to the heater in an automobile, that circulates the oil from the tube housing and exhausts the heat. However, it is advisable to have a thermostatic switch on the tube so that, should everything go wrong and the housing get too hot, the system will shut down and cool off.

The usual configuration of an ordinary phototiming system (also called automatic exposure control, or AEC) is shown in Figure 14. Because the Bucky factor of the grid is a function of patient scatter, the phototiming sensor must always be located after the grid. If the phototiming sensor is behind the cassette, variations in cassette attenuation become important and must be compensated for. Cassettes of equal sensitivity may have different x-ray transmission; with a behind-the-screen phototimer, we must use cassettes that have the same beam attenuation and sensitivity. Another problem is that the cassette absorbs energy as a function of the thickness of the patient; the patient serves as a beam filter preceding the cassette. It may be necessary to automatically compensate exposure time as a function of this change of energy distribution reaching the phototiming sensor. If we put the phototiming sensor in front of the cassette, there is no problem of energy change of the transmitted radiation through the cassette, and we have only to select cassettes of equal sensitivity. There are exceptions; an example would be an automtic film changer where you have only one pair of screens for all exposures. Most phototiming sensors will not match the screens because the energy absorbed by the phototiming sensor may be different from that absorbed by the screens. The classical, old-time phototimer had a very thin screen that glowed and a light pipe or channel that carried the light to a photomultiplier tube which terminated the exposure. Another type of sensor is a simple ion chamber where the gas is ionized; this ionization is then detected by an electrometer circuit. The sensitivity of the ion chamber is increased by making the side walls of some heavy metal photoemitter. Some of the chambers today use a very thin layer of lead on the walls. Semiconductor depleted junctions can also be used as photoconductors to terminate the exposure. One company formerly used the photocathode current of the image intensifier over the spot film device to terminate spot film exposures. The drawbacks to that system are obvious.

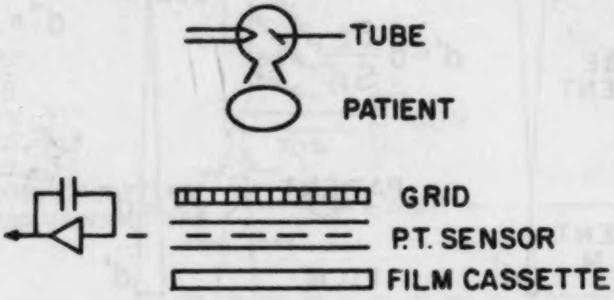

Figure 14. Phototiming system in typical configuration.

Single phototiming sensors can be used in the center of the field; this is adequate for some types of spot filming. The preferred technique is to use three sensors, two in the chest field and one behind the spine so that the operator selects the area of interest to terminate the exposure. The better systems allow one to parallel the sensors with the gain adjusted so the net exposure will be the same for one, two, or three sensors, and the net film density will be the same. Among the problems in any phototimer is that of short time compensation. The capacity of the x-ray tube power cables is about 3,000 pf each; thus, in the simple act of charging and discharging the cables, about 2 mAs may be stored. This means that there will be difficulty with very short exposure times and the cable system can dump this charge into the x-ray tube. One way around this difficulty is to apply a "fudge" factor to differentiate the output signal of the phototiming pickup and add that to the output of the sensor itself. Thus, a very fast rate of change will terminate that much earlier to provide a certain amount of short time compensation.

We must also concern ourselves with spectrum compensation. If the phototimed exposure is very short, patient thickness must be relatively small. For these patients, output through the patient will not have the usual energy distribution, so we must design the phototimer to terminate the exposure even earlier by applying an additional compensation factor.

A backup timing means is essential. The BRH allows 600 mAs as the absolute limit to phototiming chest systems, but the system should never go that far. The choice of backup time should be a function of the particular examination being conducted so that if all else fails, including the phototimer, the exposure time will not be more than, perhaps ten times normal. All phototimers should have their operation checked under varying kVp and varying area and thickness of the phantom. Figure 15 shows a phantom and penetrameter set that is available from several companies and useful for checking phototimers. You take an exposure of one block plus a 1 mm copper sheet at one value of kVp, an exposure with both blocks and a 1 mm copper sheet at the same kVp, with one block and the copper at some other kVp and both blocks and the copper at that other kVp. All exposures should be within 0.30 density units.

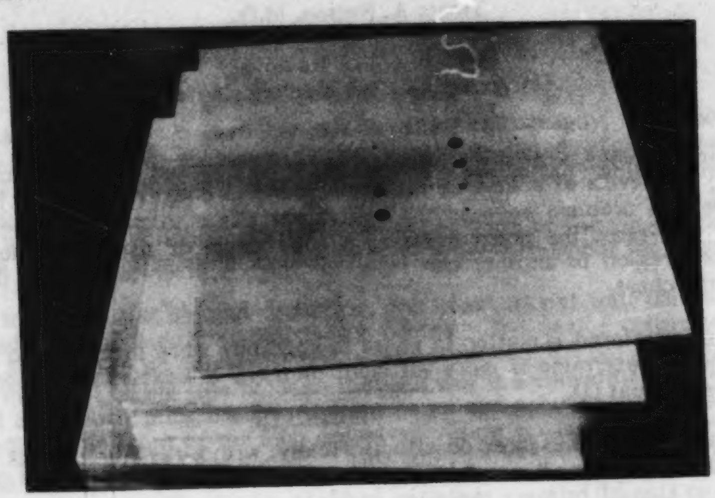

Figure 15. Phantom and penetrameter set.

Photofluorographic systems employ a very large screen and a 70 mm camera with a very fast lens that simply photographs this large fluorescent screen. From theoretical considerations alone, it can be shown that such techniques result in exposure about five times greater than conventional chest radiography, so I do not think anyone here will advocate their use.

In my view, an ideal chest system would be three phase or constant potential; it would have a medium-focus tube to avoid smearing out the images while permitting reasonable exposure times. The beam should be fairly hard and very well filtered. Collimation is crucial for minimum area, proper beam filtration, and the reduction of off-focus radiation. Focal-film distance should be chosen to provide relatively little magnification between patient and film. The film-screen combination should be fast, and should also be a function of the size of the objects of interest within the patient. System resolution will thus usually be medium to relatively low. Automatic exposure control should be effective at all points of the image area, for all exposure times, for thick and thin patients, and for varying kVp and should be tested for correct operation under all these varying parameters. System geometry and alignment are critical because none of the systems will work well if it is not installed properly.

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SCATTERED RADIATION IN CHEST RADIOGRAPHY

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In diagnostic radiology, scattered radiation degrades contrast and adds background noise to the image. The levels of scattered radiation found in chest radiography have not been considered a serious problem because the equivalent tissue thickness of the lungs is only about 10 cm. However, the chest contains a wide range of tissue thickness and cross-scatter from one region to another can result in high scatter-to-primary (S/P) ratios in some areas of the chest.

The levels of scattered radiation in chest radiography were investigated to determine their values and to determine if more efficient methods of scatter reduction would yield substantial improvement (1).

Scattered radation was measured densitometrically, with x-ray film used as the radiation detector. A lead beam stop, 6 mm thick and 6 mm in diameter, was placed on the surface of the phantom to stop all primary radiation incident upon it. Thus, scattered radiation was measured in the shadow of the lead beam stop and unscattered primary and scattered radiation in the region immediately adjacent to it. The measured optical densities were converted to radiation intensities, using a film H and D curve. All measurements were made at 120 kVp using polystyrene phantoms and a 3-phase generator.

To determine the effects of cross-scatter at an interface of two different thicknesses, we measured the intensities of primary and scattered radiation near an interface of 10-cm and 20-cm thicknesses. The absolute intensities of both scattered and primary radiation were found to be higher on the 10-cm side, even though the scatter-to-primary ratio was higher under the 20-cm side.

In chest radiography, the lung fields represent a relatively intense source of scattered radiation. The levels of primary radiation are substantially lower in the thicker regions of the chest and cross-scatter from the lungs can result in very high scatter-to-primary ratios in these areas.

In studies on a chest phantom, scatter-to-primary ratios were found to be very high in the central mediastinum, lung periphery and cardiac shadow, equalling or exceeding values observed in abdominal radiography. In these areas of the chest film, even with a 12:1 antiscatter grid, less than 50 percent of the available image contrast is being displayed. Scatter reduction techniques may, therefore, offer substantial improvements in image quality in these areas.

REFERENCE

1. Sorenson, J.A., J.A. Nelson, S.C. Jacobsen. Rotating disc device for slit radiography of the chest. (Submitted to Radiology.)

THE EFFECT OF FILM HOLDER CONSTRUCTION ON SCATTER IN CHEST RADIOGRAPHY

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While implementing an air gap technique for chest radiography, we noted that the films received considerably more scatter radiation than anticipated. Subsequent analysis indicated that the various components of the film holder contributed significantly to the scatter radiation received at the film plane. The following work was undertaken to evaluate the contribution to the production of scatter radiation by each of the various components of a commercially available film holder currently in wide use throughout the country.

In order to utilize the air gap technique, the conventional wall unit was modified to permit the addition of a patient positioning device. The device consisted of a 1/8-inch Lucite plate attached to an adjustable frame, which enabled the patient to be positioned 10 inches from the front of the film holder. The wall unit is a conventional vertical chest unit consisting of a 1/4-inch thick hardboard front panel and a Lucite paddle phototimer positioned immediately in front of the film cassette. A bracket is provided between the front panel and the phototimer to hold the grid. This study employed a 12:1 103-line-pair grid.

The object of the study was to quantitate the contribution to scatter of each component of the film holder when the system was used with the grid or air gap technique. To do this, a 1-1/4-inch Victoreen ionization chamber was positioned in the film plane. The chamber was surrounded by a 1/4-inch lead ring to insure that the only radiation measured was that falling on the front surface of the cassette which might be recorded by the film. A styrofoam jig was placed in front of the scattering medium to hold alternately a 1-1/2-inch disc and a lead plate with a 1-1/2-inch aperture; the disc, aperture and chamber were all placed on the central axis. The resulting exposures were used to calculate scatter-to-primary ratios (S/P) for the various geometries encountered. The entire wall unit was disassembled and S/P measurements were made for each component separately and for the components in various combinations. Measurements were taken at two representative kVp's, 120 kVp and 80 kVp, utilizing both grid and air gap techniques.

Using the air gap technique, the front panel and phototimer were found to contribute substantially to the increased scatter noted at the film plane. The S/P was increased by more than 60 percent at 120 kVp by the addition of the air gap positioning device, front paddle and Lucite phototimer. The Lucite positioning device did not contribute substantially to the increase in S/P. The front paddle and phototimer were found to contribute the major portion of the increased scatter. Each caused approximately a 30 percent increase in S/P over that measured with all the components removed. The 1/4-inch hardboard panel is considerably denser than the phototimer and introduces a large amount of scatter, but because of the air gap between the front panel and the film, only a small amount of this scatter reaches the film plane. The phototimer is less dense than the front panel and, therefore, causes less total scatter. Because of its proximity to the film plane, however, it does produce a measurable effect on the film.

Positioning of the grid in the conventional wall unit is less than optimal with regard to reduction of S/P. Scatter arising from the front panel of the film changer is effectively eliminated by the grid. Scatter arising from the phototimer, however, can readily reach the film. The addition of the phototimer between the grid and the film cassette increases S/P by approximately 70 percent at 120 kVp and approximately 100 percent at 80 kVp.

In view of these results, some recommendations can be made regarding construction of film holders for chest radography. If the grid technique is to be used, the S/P can be substantially improved by placing the phototimer somewhere other than immediately in front of the film cassette. One alternative would be to place the phototimer behind the cassette and a study of this possibility is currently underway. If the air gap technique is used, the front panel of the film changer can be removed since its original purpose was to keep the patient from coming in direct contact with the other components of the changer. This function is now served by the patient positioning device. Again, the phototimer would be better placed either behind the film cassette or far in front of it. This would reduce the amount of scatter reaching the film plane.

The results of this study indicate that careful consideration must be given to design and construction of the film changers used in chest radiography and that many units currently on the market are not optimized with regard to elimination of scatter radiation.

THE PERFORMANCE OF RADIOGRAPHIC GRIDS

Gerald Jensen and Carl Scheid
General Electric Medical Systems

INTRODUCTION

The performance of a film/screen imaging system is a function of many variables. The geometry, kVp, patient size, and the amount of scatter present are just a few of the items that must be considered when determining performance on a system level. Our approach to control and determine the effect of each of these variables was to develop a computer model of the system. I would like to first describe our model and show how our predicted results compare with experimental results before getting into the applications.

We begin by calculating the primary energy spectrum coming out of the x-ray tube (Fig. 1). The units of intensity are calibrated in terms of R/mAs/keV so that when the spectrum is integrated, the absorbed dose in R/mAs is obtained.

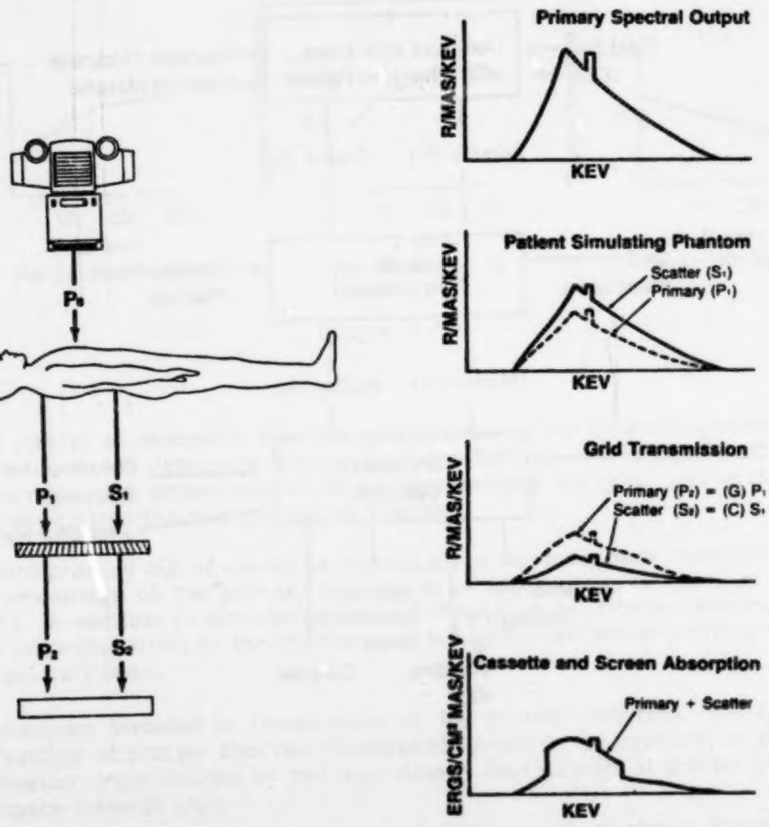

Primary Spectral Output
R/MAS/KEV
KEV
P_0
Patient Simulating Phantom
Scatter (S_1)
Primary (P_1)
R/MAS/KEV
KEV
P_1
S_1
Grid Transmission
Primary (P_2) = (G) P_1
Scatter (S_2) = (C) S_1
R/MAS/KEV
KEV
P_2
S_2
Cassette and Screen Absorption
Primary + Scatter
ERGS/CM² MAS/KEV
KEV

Figure 1

Next, by using the absorption coefficients of some low atomic number material such as acrylic, the tissue attenuation can be closely approximated. The amount of scatter present would be quite difficult to calculate since it is a function of so many variables. Therefore, we rely on the experimental results of others for this data. The P_1 and S_1 spectra both undergo some type of attenuation in the grid, resulting in reduced P_2 and S_2 spectra. Finally, these spectra reach the cassette and screen where they are absorbed; the units of the absorbed spectra are now ergs/cm^2 mAs.

Figure 2 is a block diagram showing the major inputs and outputs associated with the program. The parameters involved in generating the initial primary spectrum just discussed are the tube target angle, the kVp, and the inherent filtration of the tube and collimator. Kramer's rule is then applied, dealing with the physics involved in determining the shape and intensity of the x-ray spectrum based on the given inputs.

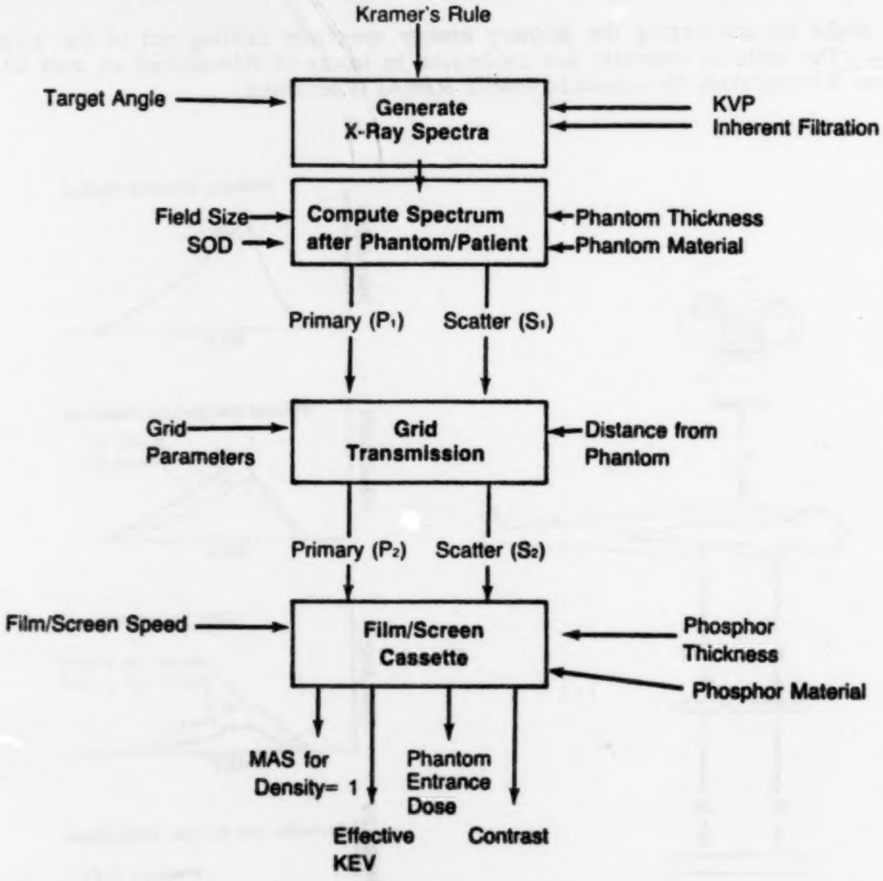

Kramer's Rule
Target Angle
Generate X-Ray Spectra
KVP
Inherent Filtration
Field Size
SOD
Compute Spectrum after Phantom/Patient
Phantom Thickness
Phantom Material
Primary (P1)
Scatter (S1)
Grid Parameters
Grid Transmission
Distance from Phantom
Primary (P2)
Scatter (S2)
Film/Screen Speed
Film/Screen Cassette
Phosphor Thickness
Phosphor Material
MAS for Density= 1
Phantom Entrance Dose
Effective KEV
Contrast

Figure 2

The primary spectrum transmitted by the phantom is a function of the phantom thickness and material. The scatter spectrum is derived from the primary by knowing the scatter fraction. The scatter fraction is the percentage of the total beam that is scatter.

For our model we use the experimental results of Motz, Dick, and Soares in obtaining the initial scatter fraction (Fig. 3). As shown, the scatter fraction is related to geometric conditions such as field size and spacing from the image plane, as well as phantom thickness. Note that it takes only 5 cm of a polystyrene phantom to have a scatter fraction of 50 percent; at 10 cm, 70 percent of the total beam is scatter.

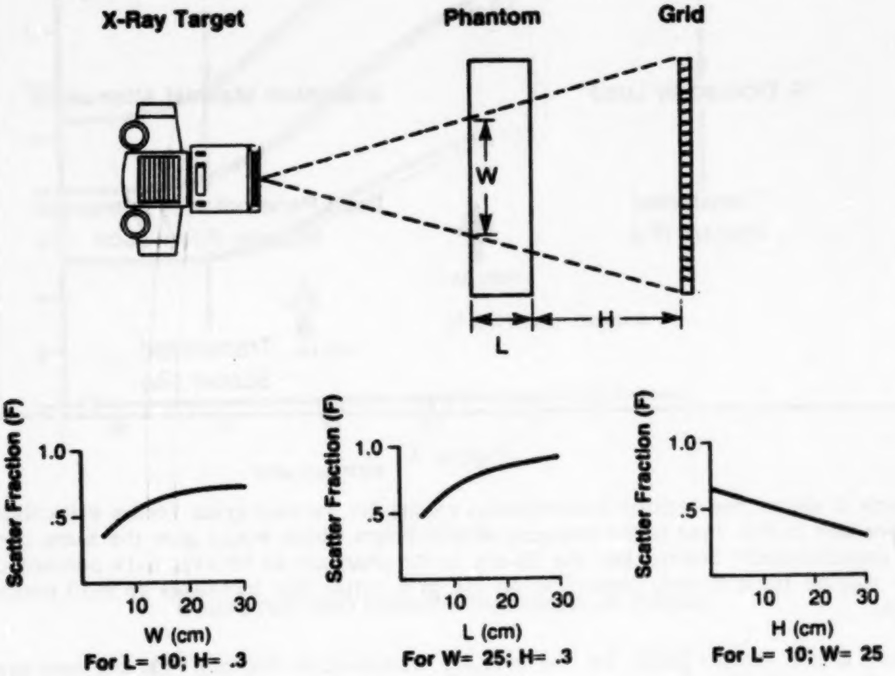


Figure 3

The next step is to determine how the grid attenuates the incident spectrum (Fig. 2). The grid properties will determine the transmission factors, while the distance from the phantom determines the actual amount of scatter reaching the grid. Let us take a closer look at what determines the grid transmission factors.

The incident primary will of course be attenuated by the interspace material of the grid. Also, some percentage of the primary is going to be blocked due to the area of the lead strips (Fig. 4). In addition to considering both of these factors, a calibration must be made to allow for nonuniformities in the thickness of the grid lines before arriving at the final transmitted primary beam.

Two factors are involved in transmission of the scatter radiation. First, there is a certain percentage of scatter that has the same direction as the primary, or at least falls within the interior angle defined by two lead strips. That amount of scatter is attenuated by the interspace material only.

The other effect deals with the percentage of scatter radiation having sufficient energy to penetrate the thickness of a lead strip. This scatter also undergoes some slight attenuation due to the interspace material.

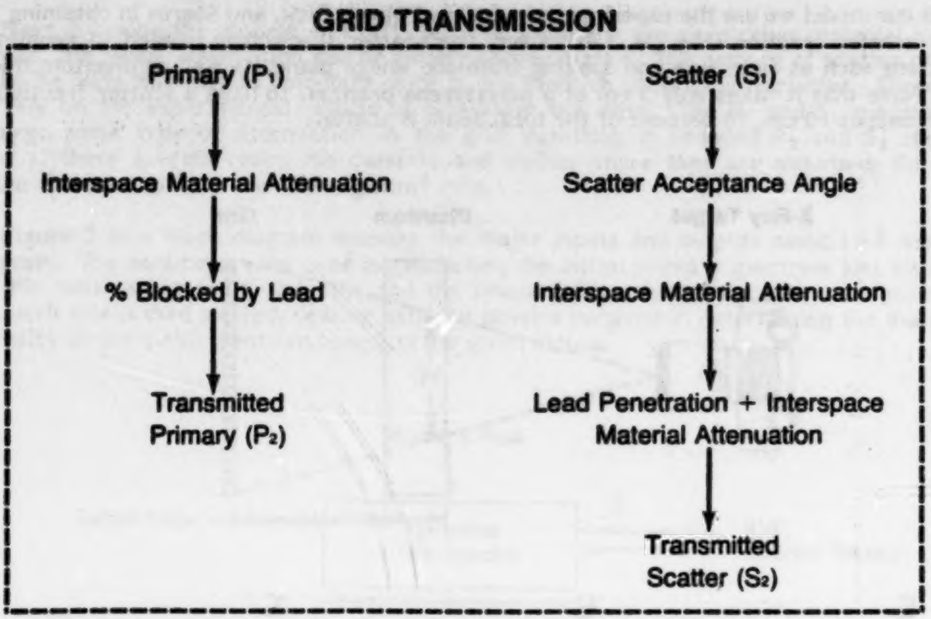

GRID TRANSMISSION
Primary (P_1)
Interspace Material Attenuation
% Blocked by Lead
Transmitted Primary (P_2)
Scatter (S_1)
Scatter Acceptance Angle
Interspace Material Attenuation
Lead Penetration + Interspace Material Attenuation
Transmitted Scatter (S_2)

Figure 4

Figure 5 shows the scatter transmission values for various grids versus effective keV. Effective keV in this case is the monochromatic beam which would give the same contrast as our polychromatic beam. For the 20-cm lucite phantom at 80 kVp, 6-14 percent of the scatter will be transmitted, depending on the grid ratio; this increases to 9-19 percent at 140 kVp.

Figure 6 is a similar graph for the primary transmission factors. On the average, the fiber grids transmit 15 percent more primary than the aluminum interspace grids, due to the properties of the material. Again, depending on the grid ratio and type used, primary transmission values range from 49-69 percent at 80 kVp to 56-73 percent at 140 kVp for the given phantom. Note that effective keV depends on both phantom thickness and kVp.

Finally, if we know the film/screen speed, phosphor thickness and material, and the primary and secondary spectra incident on the cassette, several outputs can be obtained (Fig. 2). Some of these are: mAs for unit density, the effective keV mentioned before, contrast, and phantom entrance dose. Actually, the dose at any one of the stages is available by simply integrating the energy spectrum at that point in time.

One assumption of the model is that the grid is aligned and focused to the central ray of the x-ray beam (Fig. 7). If the focal spot is both off center (v) and not at the grid focus (fo), the line of maximum transmission will occur at a distance X_f to the opposite side of center, as shown. The grid attenuation (A) at any distance $X + X_f$ can be obtained from the given equation. The shape of the density profile in a direction perpendicular to the grid lines is essentially a "haystack" that peaks at X_f.

For example, suppose we were using a 12:1 grid focused for 60 inches at 72 inches and it was 1/2 inch to the left of center. This would give the peak density 2.5 inches to the right of center and a 20 percent density falloff 3 inches to the left of center. Therefore, if lung fields were being compared, this could well be unacceptable.

To verify the model, we compared our results to some experimental results furnished by Dr. Gary Barnes (Fig. 8). Good correlation was obtained.

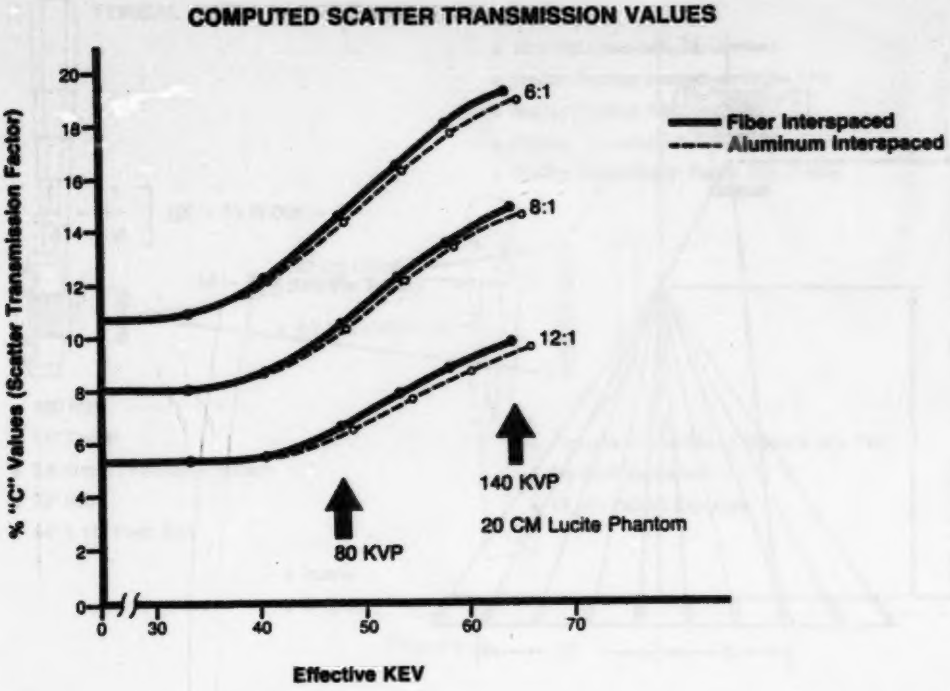


Figure 5

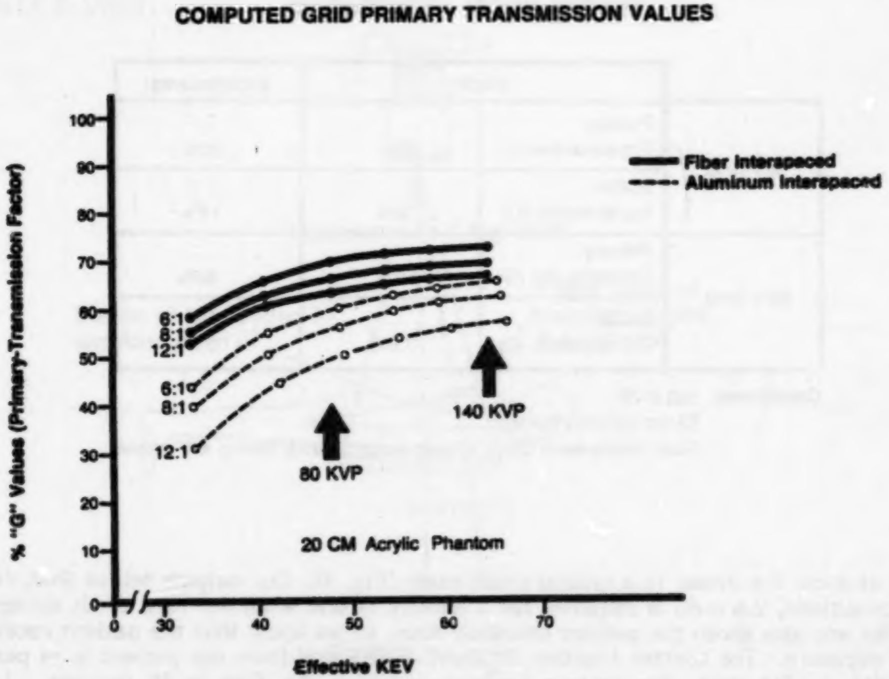
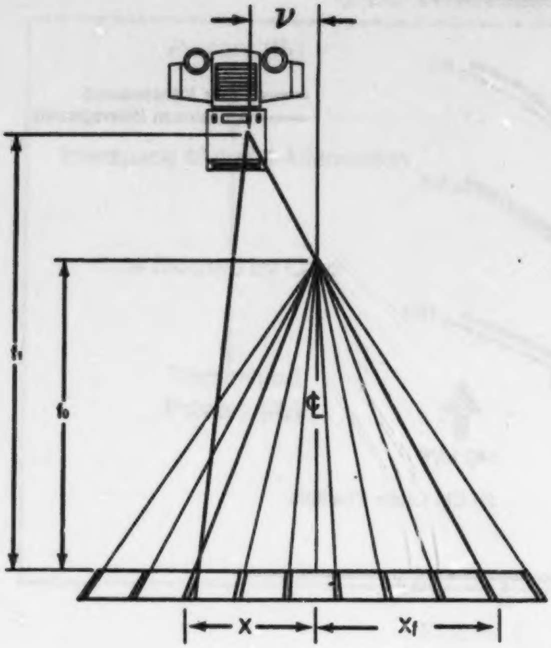


Figure 6

GRID MISALIGNMENT



$$A = 100\,R\,(X + X_f)\left[\frac{1}{f_0} - \frac{1}{f_1}\right]$$

$$X_f = \nu\, f_0/(f_1 - f_0)$$

Figure 7

SYSTEM MODEL VERIFICATION

		Model	Experimental
8/1 Grid	Primary Transmission (G)	69%	70%
	Scatter Transmission (C)	12%	14%
12/1 Grid	Primary Transmission (G)	65%	64%
	Scatter Transmission (C)	7.8%	7.4%

Conditions: 100 KVP
20 cm Acrylic Phantom
Fiber Interspaced Grids in reciprocating bucky having a 1" stroke.

Figure 8

Let us apply the model to a typical chest exam (Fig. 9). Our outputs tell us that, for the given conditions, 2.6 mAs is required for a density of one when our rare earth screens are used. We are also given the patient entrance dose, so we know that the patient received a 12 mR exposure. The scatter fraction incident on the grid from our patient is 74 percent. Even with a 12:1 grid, the scatter fraction reaching the film is 30 percent. Let us determine what a 30 percent scatter fraction means in terms of contrast.

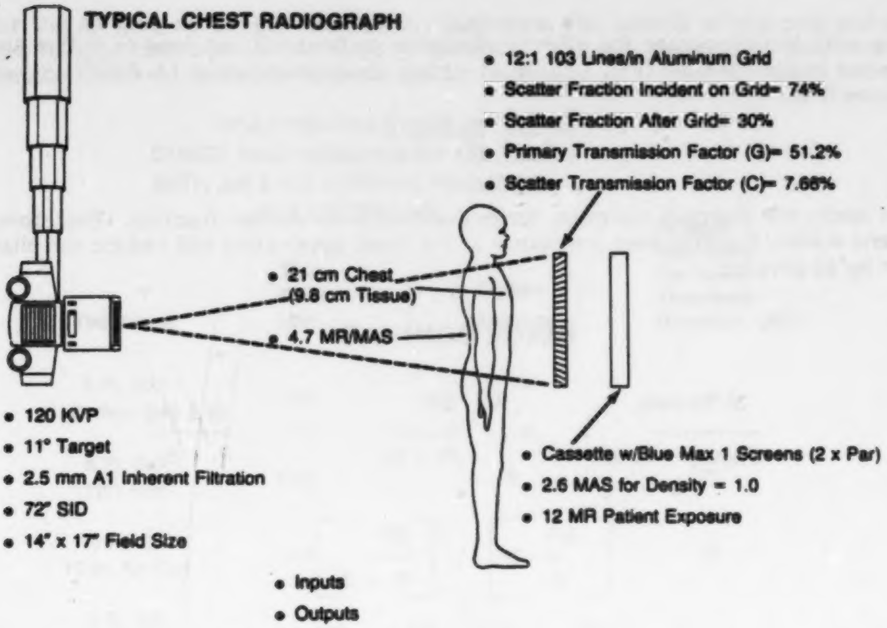
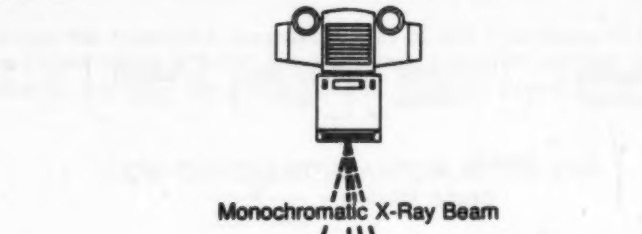

TYPICAL CHEST RADIOGRAPH
• 12:1 103 Lines/in Aluminum Grid
• Scatter Fraction Incident on Grid= 74%
• Scatter Fraction After Grid= 30%
• Primary Transmission Factor (G)= 51.2%
• Scatter Transmission Factor (C)= 7.66%
• 21 cm Chest (9.8 cm Tissue)
• 4.7 MR/MAS
• 120 KVP
• 11° Target
• 2.5 mm A1 Inherent Filtration
• 72" SID
• 14" x 17" Field Size
• Cassette w/Blue Max 1 Screens (2 x Par)
• 2.6 MAS for Density = 1.0
• 12 MR Patient Exposure
• Inputs
• Outputs

Figure 9

Consider a lesion of mass/area X_1 and some normal tissue of mass/area Xo. Coming out of the phantom are primary beams P_1^1 and P_1^0, respectively (Fig. 10). If the density and composition differences between normal and diseased tissue are small, phantom contrast is $(\Delta P_1)/(P_1)$.


Monochromatic X-Ray Beam
Lesion of Mass/Area X₁ and Attenuation μ₁
Normal Tissue of Mass/Area X₀ and Attenuation μ₀
P₁¹
P₁⁰

$$\text{Contrast} = \frac{\Delta P_1}{P_1}$$

Figure 10

If we wish to incorporate the effects of scatter on contrast, we need to define another term called image contrast (Fig. 11). If we define image contrast as (Δ P)/(P + S), we can manipulate it to:

$$\left[\frac{\Delta P_1}{P_1}\right]\left[\frac{P_2}{P_2 + S_2}\right]$$

which is really the phantom contrast, times one minus the scatter fraction. Therefore, the 30 percent scatter fraction seen previously in the chest application will reduce the phantom contrast by 30 percent.

IMAGE CONTRAST

$$\frac{\Delta\ \text{Primary}}{\text{Primary + Scatter}} = \frac{\Delta P_2}{P_2 + S_2} = \frac{G\,\Delta P_1}{P_2 + S_2}$$

$$\left[\frac{\Delta P_1}{P_1}\right]\left[\frac{GP_1}{P_2 + S_2}\right]$$

$$\left[\frac{\Delta P_1}{P_1}\right]\left[\frac{P_2}{P_2 + S_2}\right]$$

$$\left[\begin{matrix}\text{Phantom}\\ \text{Contrast}\end{matrix}\right]\left[\begin{matrix}1 - \text{Scatter}\\ \text{Fraction}\end{matrix}\right]$$

$$\left[\begin{matrix}\text{Phantom}\\ \text{Contrast}\end{matrix}\right]\left[\begin{matrix}1 - \text{Scatter}\\ \text{Fraction}\end{matrix}\right]\left[\begin{matrix}\text{Focal Spot}\\ \text{MTF}\end{matrix}\right]\left[\begin{matrix}\text{Screen}\\ \text{MTF}\end{matrix}\right]$$

Where

$$\text{Scatter Fraction} = \frac{CS_1}{GP_1 + CS_1}$$

Figure 11

If objects smaller than 2 mm are of interest, the MTF of the focal spot and screen must be considered. As shown, they become multipliers on the expression, reducing image contrast.

The scatter fraction definition shown is just a reminder of how the grid primary and scatter transmission factors are used in determining the contrast reduction seen at the image plane.

Let us return to the chest exam and consider what happens to skin dose and contrast for a couple of different techniques. Start with a perfect grid that will transmit all of the primary and no scatter, giving us a skin dose of 9.2 mR and no loss in contrast (Fig. 12). Going to a 12:1 grid increases the skin dose and reduces the contrast, as would be expected. If an air gap technique is employed, with a 10-foot SID and a 10-inch air gap, there is a dose reduction compared to the perfect grid but the contrast is also lower. The same thing holds true for the no grid, no air gap technique. Therefore, the benefit of the grid and air gap over the no grid technique is not directly evident. To make this comparison we need to consider the threshold detection of low-contrast objects.

SCATTER FRACTION IN CHEST RADIOGRAPHY AT 120 KVP WITH AN 8 CM ACRYLIC PHANTOM (9.8 CM TISSUE)

Technique	Skin Dose (MR)	[1 — Scatter Fraction]	Relative Skin Dose For Same Threshold Detection (MR)
6 Ft. SID Perfect Grid	9.2	1.0	9.2
6 Ft. SID 12/1 Grid	12.0	.70	25*
10 Ft. SID 10 In. Air Gap	6.9	.66	16
6 Ft. SID No Grid No Air Gap	2.4	.26	35.5

Figure 12

To do this we use a low contrast phantom made of four sections: bakelite, acrylic, polystyrene, and polyethylene. The hole spacing is equal to the hole diameter and the hole depth is 1 cm. The phantom is 18.6 cm deep and is filled with water.

Figure 13 shows the measured contrast values of the 4 sections of the phantom used at 120 kVp. The contrast range is from .6-4.5 percent; negative contrast values mean that the object attenuation is less than the attenuation of water as shown by the definition.

LOW CONTRAST PHANTOM AT 120 KVP

Material	Composition	ρ Density GM/CM³)	μ Attenuation Coefficient CM²/GM	Measured* Contrast
Water	H_2O	1	.21	0
Bakelite	$C_{43}H_{38}O_7$	1.3	.19	.045
Acrylic	$C_5H_8O_2$	1.2	.20	.022
Polystyrene	CH	1.05	.19	−.006
Polyethylene	CH_2	.92	.20	−.013

* Phantom Contrast $= \mu\rho X - (\mu\rho X)\text{ Water} = \dfrac{\text{CT\#}\ (\mu\rho X)\text{ Water}}{500}$

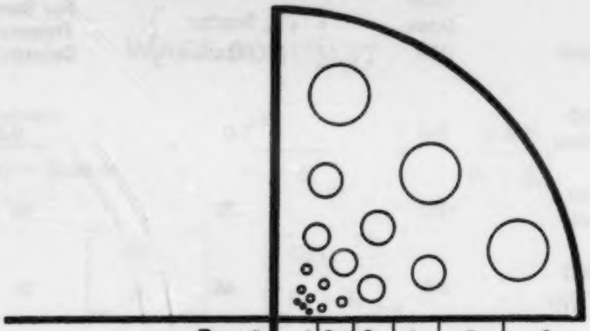
Figure 13

Figure 14 is a schematic of the water phantom showing the hole diameters used. The focal spot and screen MTF corresponding to the object size are shown because this enters into the calculation of image contrast as previously mentioned. Recall that image contrast is the product of phantom contrast, MTF, and 1 minus the scatter fraction.

LOW CONTRAST PHANTOM

Hole Depth= 1 CM

Phantom Water Depth= 18.6 CM



Row #	1	2	3	4	5	6
Hole Dia. MM	1	1.7	3	5.2	9	15.6
Holes/MM	0.5	.29	.17	.10	.055	.032
Focal Spot/Screen MTF	.86	.93	.96	.98	.99	1.50

Scatter Fraction @ 120KVP12/1 grid= .26

Figure 14

Here are some experimental results obtained with the low-contrast phantom on a system using a 12:1 grid versus a slit scanning unit using the same geometry and kVp (Fig. 15). Notice that in order for the 12:1 grid to detect the same minimum hole diameter as the slit scan unit, which we will assume removes all of the scatter, twice the phantom contrast was required. We can get the image contrast by multiplying the phantom contrast by one minus the scatter fraction, and the system MTF. Now for the same threshold diameter hole detectable, the same image contrast is obtained.

We are now getting close to reaching the contrast-dose tradeoffs (Fig. 16). A plot of the image contrast versus minimum detectable hole size just examined is shown here. Since the data are approaching the 45 degree line signifying a system limited by signal to noise, we can use the relationship:

$$\text{minimum detectable hole diameter} = \frac{\text{threshold S/N}}{\text{image contrast}\sqrt{\text{detected pho/cm}^2}}$$

Or, for the same detectability, the contrast is inversely proportional to the square root of the dose. Therefore, if we want to visualize the same hole size at 1/2 the contrast, the dose must be increased by a factor of 4.

Let us return now to the chest exam to determine what the dose/contrast tradeoffs are for scatter removal techniques versus leaving scatter in (Fig. 12). The last column is the relative skin dose to the patient for the same detectability.

THRESHOLD CONTRAST

SCATTER REMOVED

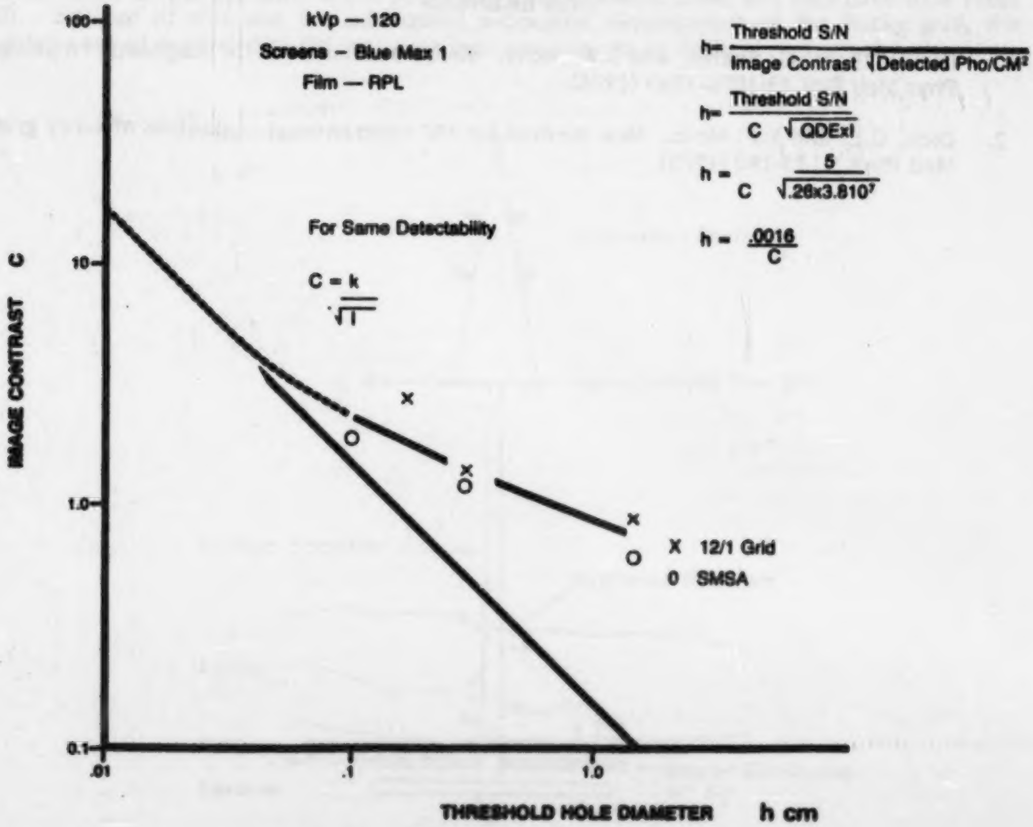
Hole Dia. (MM)	Material	Phantom Contrast	Image Contrast
1	Acrylic	.022	.019
3	Polyethylene	−.013	−.012
15.6	Polystyrene	−.006	−.006

12/1 GRID

Hole Dia. (MM)	Material	Phantom Contrast	Image Contrast
1.7	Bakelite	.045	.028
3	Acrylic	.022	.014
15.6	Polyethylene	−.013	−.009

Figure 15

THRESHOLD CONTRAST/DETAIL



$$h = \frac{\text{Threshold S/N}}{\text{Image Contrast}\ \sqrt{\text{Detected Pho/CM}^2}}$$

$$h = \frac{\text{Threshold S/N}}{C\ \sqrt{\text{QDExI}}}$$

$$h = \frac{5}{C\ \sqrt{.26 \times 3.810^7}}$$

$$h = \frac{.0016}{C}$$

Figure 16

Now the 12:1 grids show a reduction in patient dose of 42 percent compared to the no grid, no air gap technique. However, if we could remove the scatter that the 12:1 grid transmits, via slit scanning or whatever, there would be an additional dose reduction by almost a factor of 3.

We can see that air-gap techniques can also provide a decrease in patient dose. For the geometry given here, the air gap provides a 120 percent decrease in dose over the technique where no scatter is removed.

CONCLUSION

For typical chest examinations, 74 percent of the total beam is scatter coming out of the patient. Since phantom contrast is reduced by 1 minus scatter fraction at the film plane, it becomes essential that some method of scatter removal be used. However, grids are by no means perfect; therefore, additional benefits would be achieved by scatter removal systems which could better approach the ideal.

ACKNOWLEDGMENT

We acknowledge with special thanks the help of Gary Barnes, Ph.D., who provided experimental data for grid comparisons.

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PRINCIPLES OF SLIT RADIOGRAPHY

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INTRODUCTION

The use of a set of moving slits to reduce secondary radiation was initially proposed by Otto Pasche in 1903 and has the distinction of being the first scatter reduction method to appear in the radiological literature (1). Pasche's suggestion was based on the observation that "coning down," or reducing the radiation field, improved the quality of the resultant photographic plate. A moving slit device is illustrated in Figure 1 and consists of a beam-defining fore slit above the patient and a scatter-eliminating aft slit between the patient and image receptor. During an exposure, the fore and aft slits move synchronously and scan the patient. The principle of such a device is that the small radiation field defined by the fore slit produces little scatter in the area of film irradiated by the primary beam, and x rays scattered out of the irradiated volume are not imaged because of the aft slit. In order for such a technique to be effective, the width of the slit has to be small (< 1 cm at the level of the aft slit), which results in long exposure times and excessive tube loads (2). Because of this and the subsequent successful development of the Bucky grid, the use of a set of moving slits did not come into common use in medical radiography.

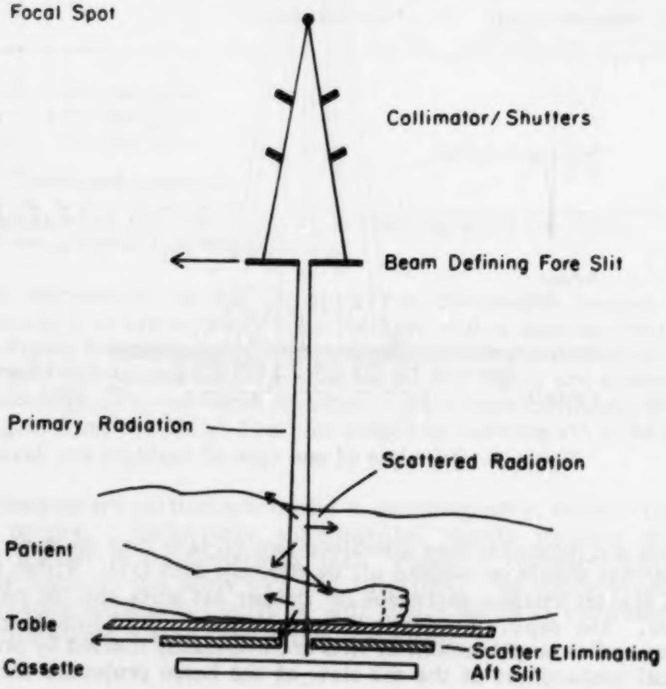

Figure 1. Principle of slit radiography.

In recent years the search for better scatter-eliminating techniques has resulted in a resurgence of interest in slit radiography (2-8). However, scatter reduction techniques using fore and aft slits did not become feasible until devices employing a multiplicity of slits were suggested (2,3,5,6,9) and high-capacity x-ray tubes and fast intensifying screens became available.

MULTIPLE SLIT TECHNIQUES

One type of multiple slit technique is illustrated in Figure 2. Rather than a single slit, an array of fore slits is positioned above the patient. During an exposure these are aligned, and moved synchronously, with scatter-eliminating aft slots (deeper than slits) beneath the patient. The high grid ratio of the aft slots (typically 8:1 or greater) eliminates single-scattered x rays arising from neighboring beams and reduces the amount of diffuse scatter that reaches the film as a result of multiple Compton events. In addition to imaging less scatter than conventional Bucky grids, such systems attenuate very little of the information-carrying primary x rays emerging from the patient. Due to the complexity of constructing aft slots, an alternative technique employed by Sorenson and coworkers is to use shallow aft slits (rather than deeper aft slots) in conjunction with a grid (10).

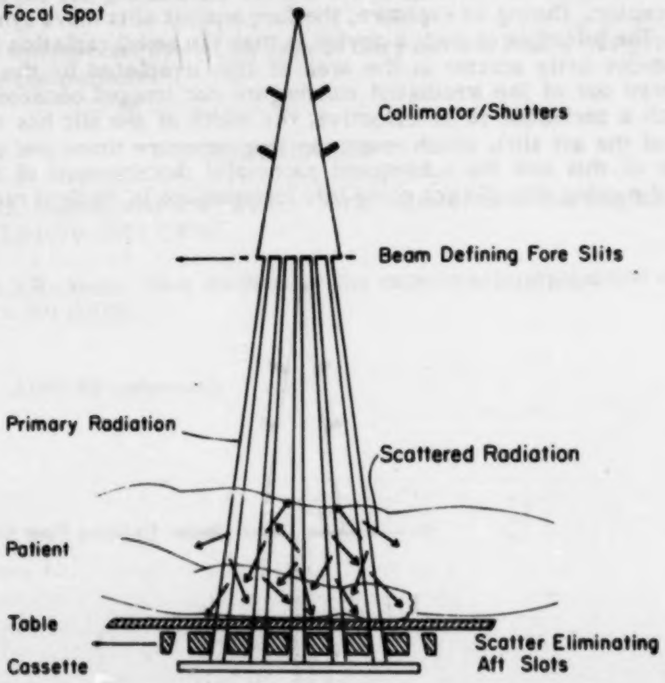

Figure 2. Principle of one type of multiple slit device.

Tube-loading and exposure-time considerations dictate that no more than 80 percent of the area of interest should be masked off by the fore slits (11). Within this constraint, the percentage of scatter imaged decreases for deeper aft slots and for narrower slits placed closer together. The depth of slots is limited by the desired radiographic magnification. The narrowness and close placement of slits are eventually limited by primary losses due to the geometrical unsharpness at the aft slots of the beam projected through the fore slits (11). When a grid is employed with the aft slits, there is less advantage in using narrow slits placed close together.

CURRENT CLINICAL APPLICATIONS

The clinical applications of multiple slit techniques that have been reported to date are listed in Table 1. This discussion is concerned with the two applications and associated devices that have been developed at the University of Alabama in Birmingham. The two devices—referred to as scanning multiple slit assemblies, or SMSA's—are shown in Figures 3 and 4. The performance of the abdominal SMSA is compared to that of conventional Bucky grids in Table 2. The SMSA transmits significantly less scatter and has better primary transmission than the best grid. This superior performance is evident in Figure 5 which compares lateral sacral spine films taken on the same patient at the same kVp. The improved image quality obtained with the SMSA led to a positive diagnosis of prostate cancer metastatic to the sacrum, which was not readily apparent on the conventional film.

Table 1. Clinical applications of multiple slit technology

Application	Year	Investigators
Abdominal radiography	1976	Barnes et al. (12) U. of Alabama in Birmingham
Chest radiography	1978	Nelson et al. (13) U. of Utah
Mammography	1978	Hogg et al. (14) U. of Alabama in Birmingham
Angiography & radiography	1979	Amplatz (15) U. of Minnesota

Table 2. Comparison of grid and multiple slit techniques

	Percent primary transmission*	Percent scatter transmission*
8:1, 80 line grid	67	11.2
8:1, 105 line grid	66	11.4
12:1, 80 line grid	60	6.4
SMSA	77	2.4

*Measured at 100 kVp, 3 phase, employing an 18 cm thick, 30 cm x 30 cm Lucite phantom.

It has been well documented in the literature that comparable images requiring less radiation can be obtained at higher x-ray tube voltages with a superior scatter reduction technique (16-20). Figure 6 compares an SMSA abdominal radiograph taken at 106 kVp with a conventional 12:1 grid film made at 76 kVp. Even at 30 kVp higher and approximately one-third the entrance skin exposure, the SMSA film has slightly more contrast. When compared side by side to the grid film, the SMSA film was judged by radiologists to be diagnostically superior.

Multiple slit techniques are particularly useful in mammography, since little or no loss of primary radiation occurs. Preliminary experimental results indicate that an SMSA effectively reduces scatter in mammography (21). Figure 7 compares conventional and SMSA mammograms made at the same kVp and filtration. The improvement in contrast with the SMSA is readily apparent; clarity of soft tissue detail is increased and a small cluster of microcalcifications is seen deep within the breast.

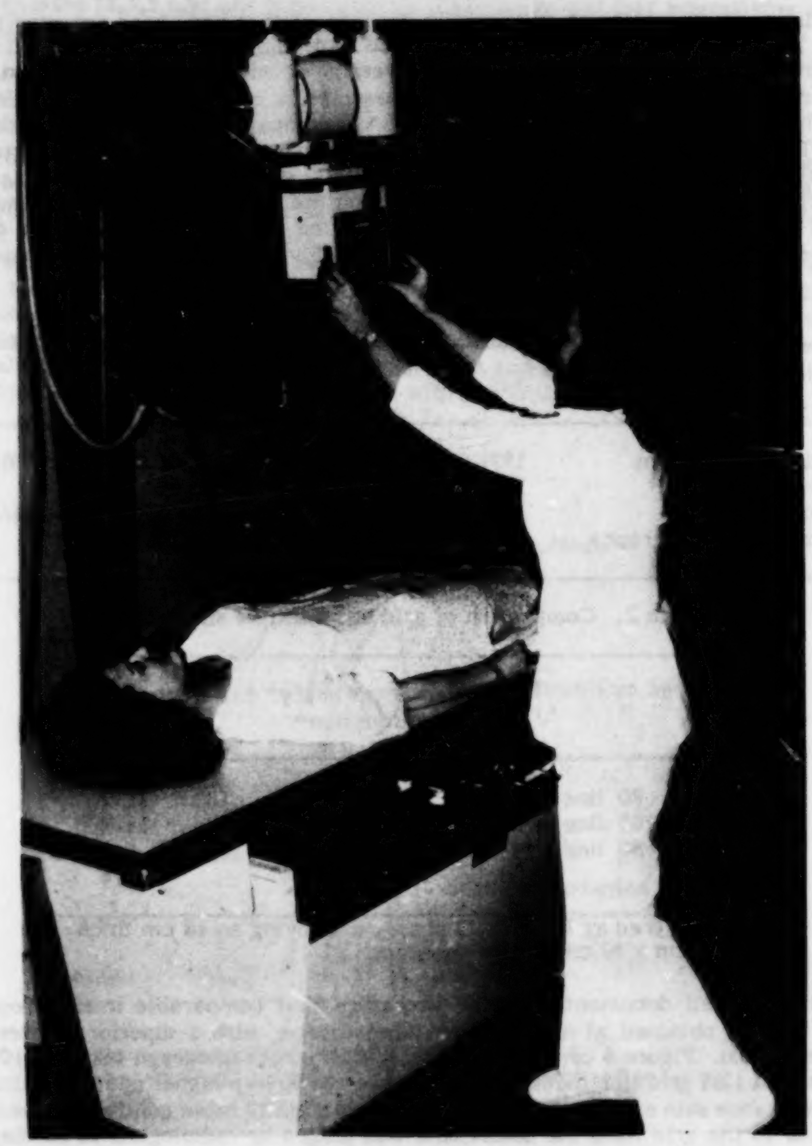

Figure 3. Abdominal SMSA. Thirty-nine fore slits and aft slots are employed. Focal spot-to-film distance is 122 cm. Fore slits, located 48.8 cm below the focal spot, are 1.63 mm wide and are separated by 6.50 mm. Aft slots are located between the tabletop and the plane of the image receptor, 117 cm from the focal spot. They are 3 cm deep, 4 mm wide and separated by 16 mm. Leading edges of adjacent slots are thus 20 mm apart. During an exposure, the area of interest is scanned by eight different beams.

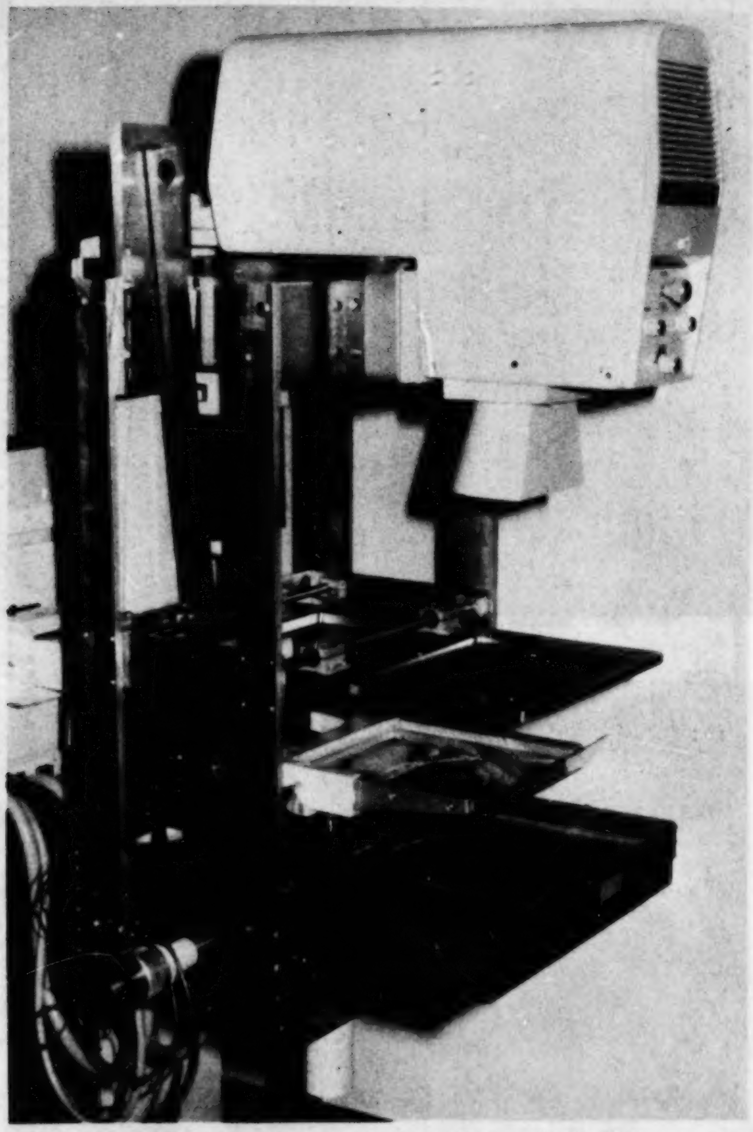

Figure 4. Mammographic SMSA. Thirty-one fore slits and aft slots are employed. The fore slits, aft slots and image plane are located at distances of 55 cm, 77 cm and 81 cm, respectively, from the focal spot. The fore slit width and spacing between slits are 2.06 mm and 5.16 mm. The aft slot width, spacing between slots and slot depth are 3 mm, 7.5 mm and 3 cm, respectively. Leading edges of adjacent slots are thus 10.5 mm apart. During an exposure, the area of interest is scanned by seven different beams.

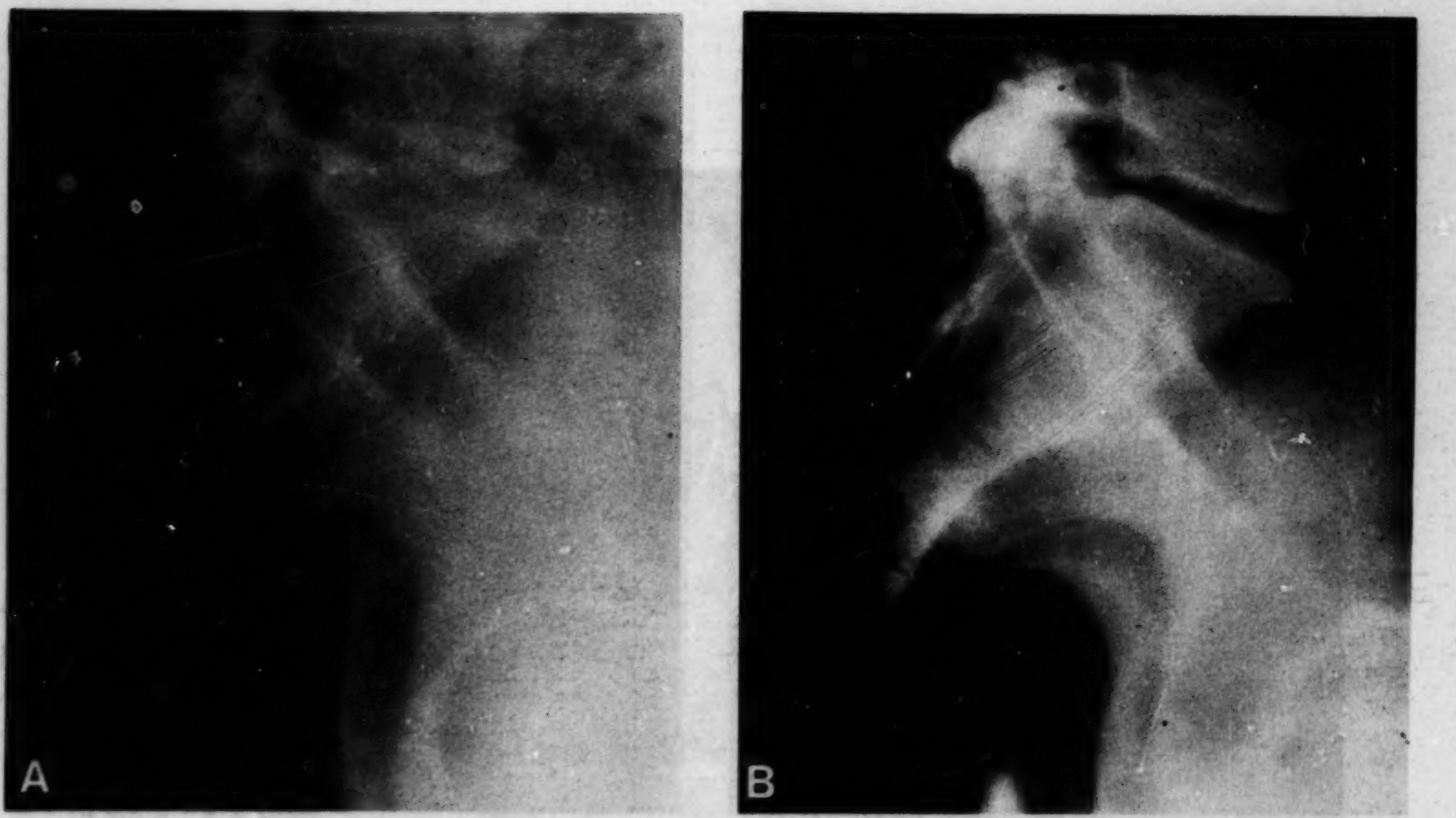

Figure 5. Comparison lateral sacral spine radiographs at 116 kVp (3 phase). Subject measured 33 cm and radiation field was 20 cm x 30 cm. (a) 12:1, 40.6 lines/cm grid using 300 mA at 0.07 sec, 102 cm SID, 520 mR entrance skin exposure. (b) SMSA using 600 mA at 1/3 sec, 122 cm SID, 760 mR entrance skin exposure. Both exposures were made employing a nominal 1.2 mm focal spot with same cassette using Kodak Lanex screens combined with Kodak Ortho-G film. The films were developed within minutes of each other in the same processor. The same generator was employed for each and the HVL's of the two source assemblies used were matched at 3.9 mm of Al at 80 kVp.

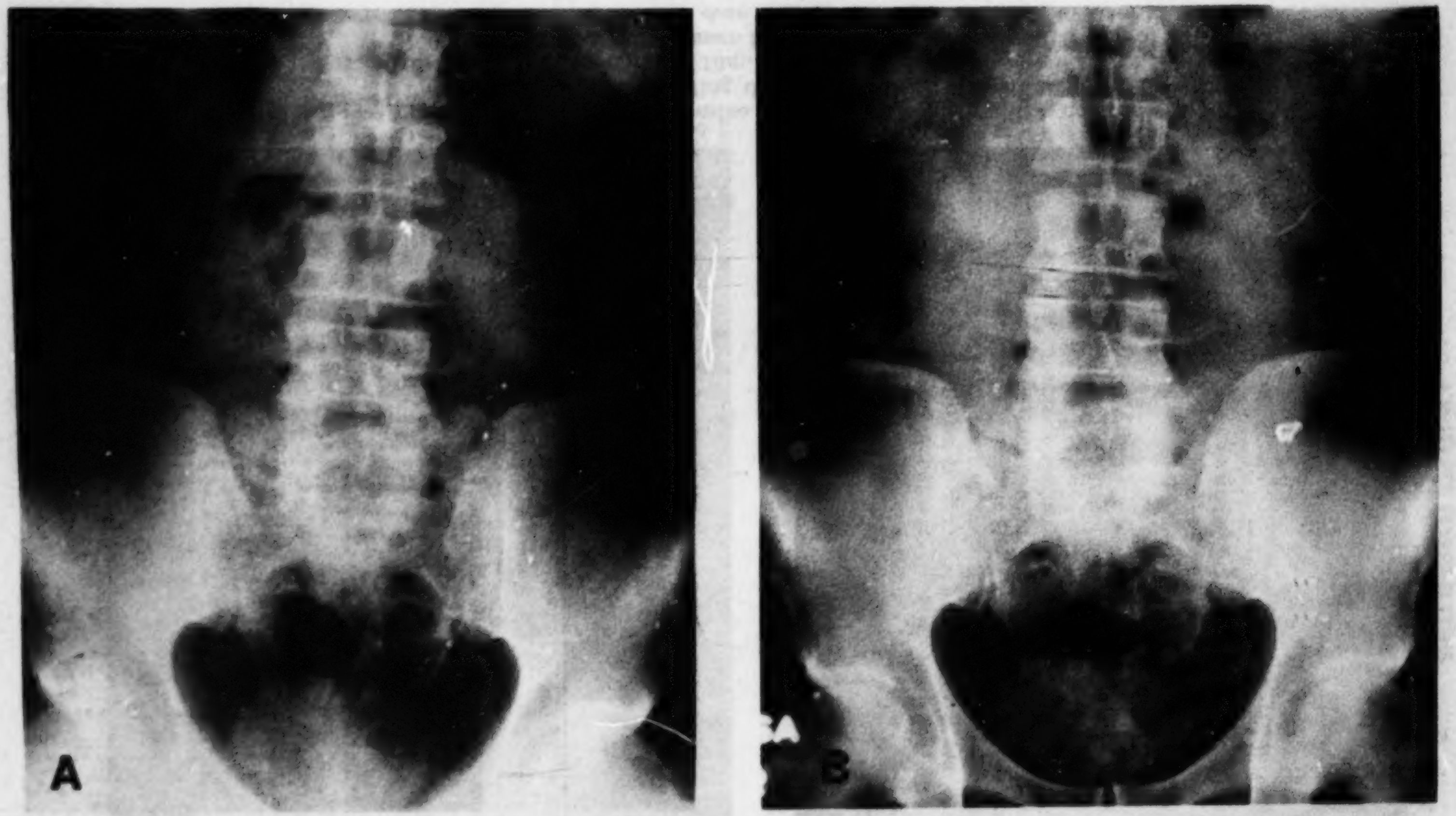

Figure 6. Comparison AP abdominal radiographs illustrating the exposure-reducing capability of SMSA. Subject measured 25 cm and radiation field was 36 cm x 43 cm. (a) 12:1, 40.6 lines/cm grid using 76 kVp (3 phase) and 600 mA at 0.15 sec, 102 cm SID, 700 mR entrance skin exposure. (b) SMSA using 106 kVp (3 phase) and 600 mA at 1/3 sec, 122 cm SID, 255 mR entrance skin exposure. Both exposures employed a nominal 1.2 mm focal spot with same cassette using DuPont Hi Plus screens combined with Kodak RPL film. The films were developed within minutes of each other in the same processor. The same generator was employed for each and the HVL's of the two source assemblies used were matched at 3.9 mm of Al at 80 kVp.

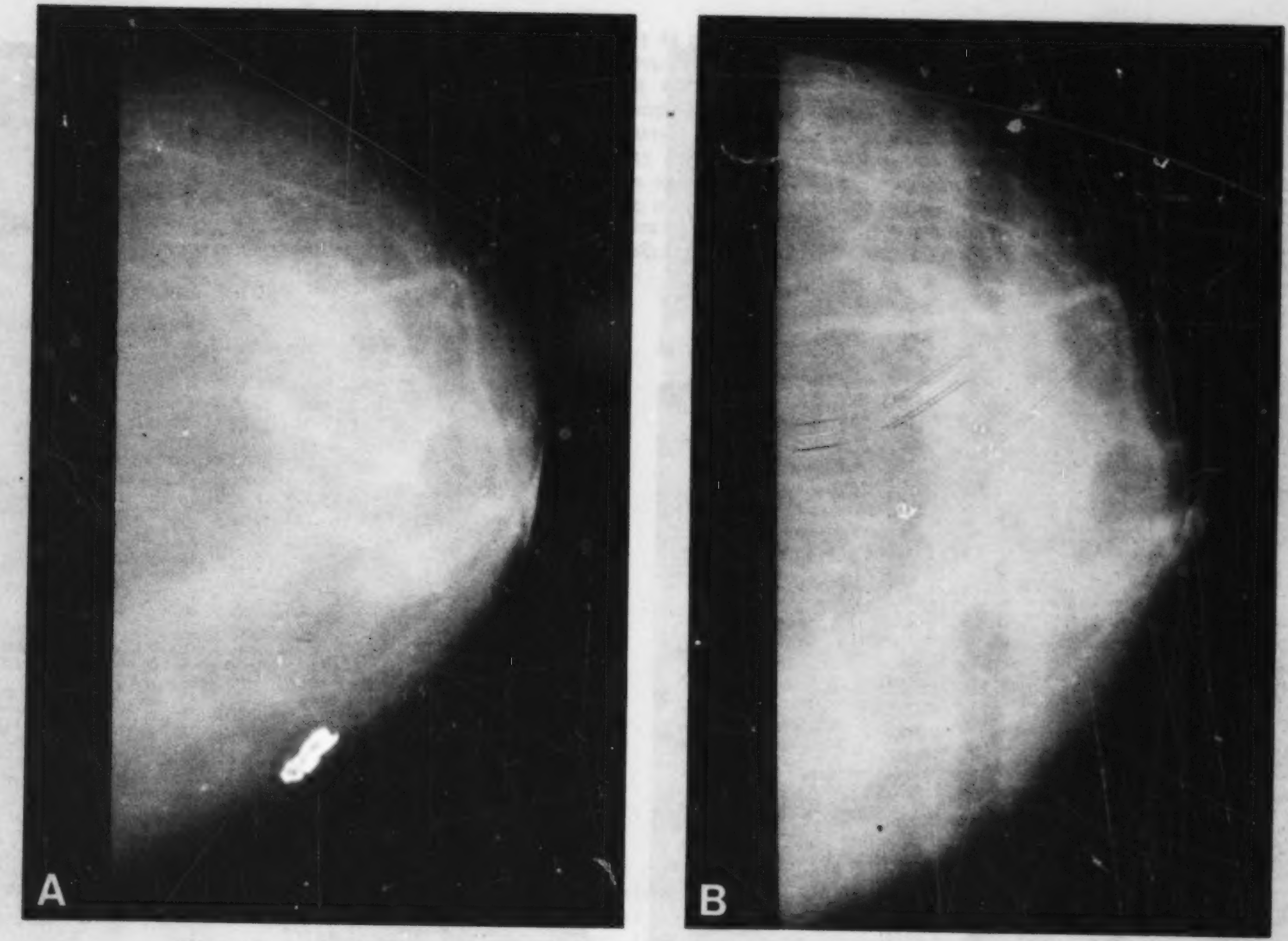

Figure 7. Comparison mammograms of a 66-year-old woman with a compressed breast thickness of 5 cm, illustrating the contrast improvement achieved through use of SMSA. (a) Conventional film. (b) SMSA film. Both were made at 32 kVp and 200 mA utilizing a molybdenum target/filter combination (0.27 mm of Al HVL). Exposure time and skin dose values are 1/4 sec and 420 mrad for the conventional film and 2 1/4 sec and 730 mrad for the SMSA.

FUTURE POSSIBILITIES

It is apparent from the preceding discussion that multiple slit technology is a practical and dose-effective method of reducing scatter. Furthermore, the clinical potential of such technology has yet to be fully explored. It is likely that at least in the immediate future such applications will be limited to a fixed tube and image receptor geometry. The experimental and clinical results discussed were obtained on some of the first clinical prototypes built, and significant improvements can be made. If wavy rather than linear slits (22) are used along with deeper aft slots, it is feasible to construct a device with a 40-percent field of coverage, a scatter transmission of 1 percent, and a primary transmission of better than 90 percent. Such a device would virtually eliminate scatter from the largest patients encountered in medical radiography.

ACKNOWLEDGMENTS

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SLIT RADIOGRAPHY OF THE CHEST

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The objective of slit radiography is to improve contrast by decreasing the amount of scattered radiation recorded on the radiographic image (1-3). In previous studies we determined that a slit device creating approximately 1-cm-wide slit x-ray beams and used in conjunction with an anti-scatter grid could reduce scatter levels significantly relative to what could be acheved with grids alone (1). We also noted that air-to-soft-tissue contrast could be improved substantially, even at relatively high kilovoltages. Therefore, we have designed and constructed a slit-radiography device for application to the chest to improve detection accuracy for nodules and visualization of other soft tissue structures in the chest and lungs (4).

Briefly, the device consists of multiple slits in a pair of rapidly rotating discs. One disc is located between the x-ray tube and the patient; the other is between the patient and the x-ray film cassette. Typically, a 6:1 grid is positioned in front of the film cassette to achieve maximum scatter reduction. Focus-to-film distance is 10 feet.

The x-ray tube is located on the common axis of rotation of the discs to project the shadow of the front slits accurately onto the slit openings in the second disc. There are 80 slits per disc. The larger disc has a diameter of about 1.5 m; its individual slits are 1-cm wide and are separated by 5 cm center-to-center at the outside diameter. The rotational speed of the pair of discs is variable, up to 10 RPS, which is sufficiently rapid to eliminate visible images of the slits in exposure times as short as 1/60 second. Typical times for PA views on adult patients are 1/30 to 1/20 second using Kodak XL film and Quanta III screens.

Measurements have shown that the slit device with 6:1 grid decreases the amount of scattered radiation recorded on images of a chest phantom by factors of 2 to 3, with the greatest reductions seen in retrocardiac and central mediastinal areas (4). However, measured (and visualized) improvements in image quality in these areas are less dramatic than expected. This is because, on conventional films, most of the blackening in retrocardiac and central mediastinal areas is due to scattered radiation. When this radiation is eliminated, the film is underexposed in these areas, even more than on conventional films. The gain in x-ray contrast is thus partially negated by the loss of film contrast.

There does not seem to be a film type available at present that can avoid this problem of limited latitude and contrast in chest radiography, which is made even more severe by the elimination of scattered radiation in slit radiography. We have been investigating optical masking techniques in an attempt to solve this problem. Preliminary results have been very encouraging (5). Currently, we feel that slit radiography of the chest has substantial promise but that its practical application will require solution of the film latitude/contrast problem.

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CLINICAL EXPERIENCE WITH SLIT RADIOGRAPHY OF THE CHEST

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Using the device described in the preceding paper, we have begun a series of clinical studies to compare slit radiography of the chest with routine techniques. The protocol has been approved by the Human Subjects Committee and the Isotope Committee at our institution. In general, we have obtained matched pairs of PA and lateral chest films using both routine and slit techniques. Film densities, controlled by phototimer settings, have been matched as closely as possible so that the photographic qualities are similar. Precise patient positioning has been relatively easy to achieve using the standard collimator light, a chin rest, and toe holders.

We have now filmed 22 patients, all chosen for evaluation of lung metastases. No unbiased comparison of films has been performed so far. In an open, side-by-side comparison of routine versus slit films, there was a subjective improvement in the clarity and perception of lung nodules. However, no statistically significant improvement has yet been demonstrated. A larger scale, double-blind study is being undertaken to determine whether such a statistically significant improvement exists.

Because the slit exposure must be recorded on faster film, slit films are more mottled than routine films (1). In thin patients, scatter reduction of the slit system reveals mediastinal structures and abnormalities better than routine films. In normal adults, however, slit films demonstrate underexposure of the light areas of the film due to the pronounced scatter reduction (1).

In an effort to overcome the problem of the underexposed mediastinum, a technique for optical masking has been developed (2). In initial clinical trials, optically masked films show a dramatic diagnostic advantage over unmasked films.

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DISCUSSION

DR. MILNE: Dr. Nelson, that is very exciting work on slit radiography of the chest. I think you may be somewhat disappointed at present that there has not been a more dramatic improvement. The breast and the abdomen work was certainly quite dramatic and I would have anticipated, as you did, that the chest work would be more dramatic. Still, I think it is the right thing to do. At the present stage, when you are struggling to get that mediastinal image back, have you looked at a normally penetrated view of the mediastinum and then, on the same patient, at your post-masked appearance of the mediastinum? Is it worth all that effort to bring back the mediastinum, particularly at 3 times the skin dose, or would it be more sensible to use your mask technique for the lungs only and a straightforward overpenetrated view for the mediastinum?

DR. NELSON: That is an excellent question, and it is also one of the first ones that came to my mind. One thing we must have is an anthropomorphic phantom--so that we know what the truth is, so that we can do a decent study. This is really hard to achieve with patients. When I do discover something on these patients prior to whole lung tomographs, then the oncologist says, "Aha! Now if you want whole lung tomographs, you pay for them." At this point I am put in the rather difficult position of proving what the truth is. In fact, on a couple of these patients, Dr. Milne, we have gotten overpenetrated views. They must have been obtained at too high a kVp or something, however, because although they are overpenetrated we are not seeing the mediastinal structures clearly. The film I showed you yesterday demonstrates a golf-ball-sized retrocardiac lesion which could not be seen on multiple overpenetrated views, including obliques of the chest, but is well demonstrated by the mask technique.

DR. SARGENT: I object to the use of the term "overpenetrated." Either you penetrate sufficiently or you do not. It is a term used indistinguishably from "overexposed." It is simply inaccurate.

DR. NELSON: I am not a physicist, but I will admit that is a kind of clinical jargon we probably should explain to our physicist colleagues. Do you have a good definition of "overpenetrated" view, Dr. Milne?

DR. MILNE: "Overpenetrated" in comparison to the lungs.

DR. NELSON: That is when we do not worry about losing lung detail and we go ahead and increase the exposure so that we can see the spine easily through the mediastinal structures.

MR. SCHEID: You previously mentioned unsuccessful attempts to use a tunnel filter or some other filter. Can you describe what the problem is?

DR. SORENSON: We attempted to make a filter for a Rando chest phantom. After spending a lot of time making it, we found we could still see the shadows of the filter on the images of the chest phantom. We then investigated the optical masking technique, which gave such an improved appearance that we have not returned to the external and wedge filtering techniques. I think they possibly could be developed but we were not too successful nor were we very pleased with our initial attempts.

DR. NELSON: The slide that Dr. Sorenson showed of the man with the very large right lower lobe and left upper lobe infiltrate is a perfect example of why a wedge filter will not work in a clinical situation. We also get a lot of people with scoliosis, a lot of postoperative patients, patients with mastectomies, for instance, for whom the wedge filter will sometimes actually work against you clinically. One of the delightful things about the masking idea was that you tailor make the thing for every patient and for his pathology, so the resulting film is hand tailored.

DR. MUNTZ: If you use the maximum film-focus distance for your tube loading without your slits and then use a huge air gap with reasonable magnification, how much scatter reduction do you get compared with your slit device?

DR. SORENSON: We have not looked at huge air gaps, but we have evaluated up to 10-inch air gaps. Scatter radiation with the slit is reduced by a factor of 2 or 3 compared to the 10-inch air gap.

DR. MUNTZ: What is the loading increase for your slit?

DR. SORENSON: About a factor of 7 with the slits--a factor of 5 because the slit assembly blocks 80 percent of the x-ray field, plus the fact that when you eliminate scatter radiation you have to increase the primary to get adequate film exposure.

DR. KELSEY: I am interested in how long the optical technique takes. How long is the patient standing there before you are ready to take the second film?

DR. SORENSON: It takes about 2 minutes to obtain an adequate mask film with the diffusing plates and so forth, so you have to take the scout film, develop it, expose it for 2 minutes in a copy box, develop that, load the cassette--probably 10 to 15 minutes total.

DR. RAVIN: Dr. Sorenson, could you comment on the validity of measuring scatter-to-primary ratio using films as opposed to direct measurements?

DR. SORENSON: What is a direct measurement?

DR. RAVIN: A measurement with a Victoreen chamber, for example.

DR. SORENSON: The techniques are essentially the same except that in one case the film is used for a radiation detector--other people have used sodium iodide or ionization chambers. In theory, they should all provide the same results. Film is very difficult to use when you have very small scatter-to-primary ratios because you get way down on the toe of the H and D curve where it is difficult to measure densities accurately. However, in our studies, we have never gotten below a scatter-to-primary ratio of about 0.2, so we are up a little way on the H and D curve. On the basis of comparing our measurements to the measurements of others under similar geometries, doing reproducibility studies, and that sort of thing, we think we are within about 10 percent of absolute values.

DR. MOORE: Has anyone here compared your images to those you get by making Log E-tronic copies of conventional chest films?

DR. SORENSON: No, we have not. This technique is only about 2 months old, so we have not gotten very far in our comparison with other techniques.

DR. MUNTZ: In your data you present scatter-to-primary ratio, do you not?

DR. SORENSON: Yes.

DR. MUNTZ: That ratio goes through wild gyrations. Is it not interesting to think about just in terms of scattering intensities, because most of those gyrations are due to differences in primary attenuation?

DR. SORENSON: Correct.

DR. MUNTZ: I did some quick calculations and found that your scatter was constant within 20 percent, over your chest films.

DR. SORENSON: That is true. In fact, looking at uniform thicknesses of scatterer and absorber, scatter decreases slightly with increasing thickness. It increases when you add the first centimeter or so of scattering material, then stays fairly constant, then decreases slightly. The thing that really changes is the transmitted primary radiation intensity and that is what changes the scatter-to-primary ratio.

MR. PLEWES: In your analysis of scatter rejection and tube loading, have you compared your slit device against a pair of criss-crossed 12:1 grids?

DR. SORENSON: No. We have no experience with that.

MR. FIELDS: Have you considered the use of a double film cassette rather than masking?

DR. SORENSON: Yes, I believe the idea of using two different film/screen combinations was described yesterday. This is basically the same idea as taking a well penetrated view and then a lung view. I cannot really explain why it has not been found acceptable. I think presenting all the detail on one film provides a display that is very attractive and appealing to most people. There may be areas between the lung and mediastinum that can be seen better on the masked view than on either the well penetrated or conventional lung views.

DR. CHARBONNIER: Dr. Nelson, you mentioned 600 mA at one-fortieth of a second, or 15 mAs, without a mask. With the mask technique, you could triple the exposure to about 45 mAs. At 140 kV, that sounds like a tremendous amount of radiation. What is the patient dose?

DR. NELSON: It depends on the film/screen combination. That would be the dose we would get if we used the same film/screen combination but, as I indicated, we have to go to a very fast film/screen system in order to keep exposure time short--within one-twentieth to one-thirtieth of a second. When we combine the mask with the fast film/screen system and the slits, the entrance dose is about 30 millirem.

DR. CHARBONNIER: With the XR Film and Quanta III screens you use, the noise must be atrocious.

DR. NELSON: It is a very fast system and you can look at the films we have on display. As I say, they may not be clinically acceptable. The way to evaluate this might be to use phantoms, with longer exposure times, to see the results with a good film/screen system. We could then consider how to shorten exposure time and deal with patient dose problems if the improvement appears worthwhile.

DR. CHARBONNIER: Do you use two masks?

DR. SORENSON: No, we used only one. We did one experiment in which we tried two masks--one in front and one in back of the film. In this single experiment we succeeded in reversing the densities: the lungs became "white" and the mediastinum became black.

The use of two masks requires very careful attention to the exact optical densities of the masks. When you use only one you have a self-limiting feature. You are only working on half of the image-forming light and the films still look like chest films when you finish.

AIR GAP TECHNIQUE

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When x-ray photons strike an object, secondary radiation is produced in all directions. Utilizing the ordinary geometry for clinical roentgenographs--with the object being immediately opposed to the x-ray cassette--a large percentage of this radiation reaches the screen and produces light, exposing the film. This scattered radiation degrades detail. In the usual method for making roentgenograms of the chest, it is necessary for a grid to be interposed between the patient and the x-ray screen. This grid permits only relatively parallel x-ray photons to interact with the screen. In addition to absorbing the angled secondary radiation, the grid also absorbs a portion of the parallel beam which has traversed the patient. It is, therefore, necessary to increase the amount of radiation to which the patient is exposed when one is utilizing a grid.

With the air gap technique, the patient is removed from the x-ray cassette, creating a gap. A portion of the nonparallel radiation does not impinge upon the x-ray screen, since its angle is such that it is diverted away from the screen and cassette. Since a grid is not necessary when utilizing the air gap technique, the amount of patient exposure to ionizing radiation is considerably less than that necessary utilizing standard technique and a grid.

For several years we have been utilizing a 10-inch air gap for routine chest radiographs. To minimize penumbral unsharpness, we use a 10-foot anode-to-film distance. We have determined experimentally that the air gap of 10 inches removes sufficient scatter to provide a radiograph of the proper contrast.

We utilize this technique in association with rare earth screens and matched film -- (Lanex and Ortho G film). The focal spot currently being utilized is .9 x .9 mm. A .3 mm focal spot is also available and is utilized for smaller patients. The speed of the rare earth system permits us to utilize exposures ranging from 6 - 8 to up to 20 milliseconds at a low mA setting (200 mA). This low mA minimizes "blooming" of the focal spot. We utilize 130 kv, 3-phase, with 2-1/2 cm of aluminum filtration. An automatic chest changer, incorporating the vacuum principle to obtain excellent film/screen contact, is fed directly into a processor.

The advantages of this technique include minimizing the patient dose, and obviating the need for a grid or a bucky. The exposure times are short and could be shorter if a higher mA station were utilized. Since we do not have forced extinction in our system at this time, we have chosen to use a lower mA station and slightly longer exposure durations to minimize exposure error. Using the low mA permits us (1) to use small focal spots with a consequent geometric advantage, and (2) to minimize blooming. The roentgenograms have a wide contrast latitude which we consider advantageous for examination of the chest.

A disadvantage of the system is that approximately 8 percent magnification results from using the air gap. This means that the radiographic image of large patients will frequently not fit on a 14 x 17 x-ray film in the upright position, and it may be necessary to expose two films in these individuals. A 16 x 16-inch format would be preferable, or a 14 x 17-inch format with the long dimension horizontal. In addition, a room of approximately 18 feet in length (preferably 20 feet) is necessary to obtain the required distances. When comparing films made by the air gap technique with those made using conventional technique, it is necessary for the roentgenologist to bear in mind that the air gap technique does produce some magnification, and films made with this technique should be identified as such.

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PLAIN FILM TOMOGRAPHY FOR THE EVALUATION OF PULMONARY ABNORMALITIES

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Plain film tomography is helpful in determining whether a pulmonary abnormality is present when an indeterminate density is seen on a chest x ray. If a pulmonary abnormality is present, plain film tomography is the most accurate way to determine whether the lesion is calcified. Plain film tomography also gives information about the shape of the lesion, which aids in determining whether it is a neoplasm, granuloma, scar, or pleural plaque.

AP and oblique tomography of the hilum of the lung provides valuable information. It is a very accurate way to determine whether the hilum is normal or abnormal. In addition, tomography of an abnormal hilum usually makes it possible to determine whether the abnormality is due to a mass, an enlarged pulmonary artery or vein, or an enlarged lymph node.

Plain film tomograms of lesions in the lung apex are helpful in determining whether a density in the lung apex is due to thickened pleura or to a neoplastic mass. Plain film tomograms also are frequently helpful in determining whether there is involvement of the vertebra by a mass in the lung apex. However, plain films are more accurate than plain film tomograms in determining whether there is rib involvement by masses in the lung apex.

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ASBESTOSIS AND PNEUMOCONIOSIS SCREENING

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Today's energy crisis and tehcnological advances have reemphasized the importance of the dust pneumoconioses, particularly coal worker's pneumoconiosis, silicosis, and asbestosis. Without a tissue biopsy, the diagnosis of pneumoconiosis during life can only be made with the use of a chest radiograph and a lifetime occupational history.

The most desirable chest radiography for the study of the pneumoconioses involves techniques or procedures which will show the finer lung structures in greatest detail. While it is highly desirable to visualize the mediastinum equally as well as the peripheral pulmonary structures on a single routine screening film, this is usually not possible except in extremely thin patients. With routine screening procedures the radiologist must compromise, and a film in which the vertebral bodies and the larger pulmonary blood vessels are faintly visualized through the overlying cardiac shadow is most acceptable. In addition, the costopleural junction along the lateral thoracic wall must be visualized with sufficient detail.

Films with a relatively long gray scale are more desirable than films of very high contrast. In addition, it has been shown that overexposed films result in the disappearance of small opacities, while underexposed films will be interpreted (particularly by inexperienced readers) as having more disease than is actually present. High kilovoltage technique (110 - 140 kV) with a stationary grid is most desirable for routine screening of patients. This technique has the greatest latitude, with a large margin for error, and produces radiographs of maximum detail with minimum variation in density and contrast.

The radiographic changes associated with all pneumoconioses may be classified using the ILO 1971 classification. The classification does not attempt to define pathological entities but has great value in reporting the type and extent of the radiographic changes and in describing any progressive changes. The small opacities which are found on the radiograph are either small rounded or small irregular opacities. The small rounded opacities which are commonly associated with coal and silica are found in the upper areas of the lung. They are classified according to diameter: p (1 - 1.5 mm), q (1.5 - 3 mm), and r (3 - 10 mm). Small irregular opacities which are more commonly associated with asbestos occur in the basilar portions of the lung and cannot be measured. They are classified as s, t, or u depending on the coarseness: s opacities are fine hair-like shadows, t opacities are somewhat more coarse, and u opacities are very coarse blotchy shadows. Any opacity which is larger than 1 cm in diameter is called a "large opacity." Large opacities may be associated with progressive massive fibrosis and lung tissue destruction as well as severe emphysema.

The general contamination of the urban atmosphere and industrial exposure of workers to asbestos dust has only recently become a matter of great public concern. The major cause for failure of various public health agencies to control asbestos dust exposure is the long latent period (often 20 years or more) between the beginning of work exposure and the appearance of overt disease. The disease now being discovered was commonly initiated by industrial exposure during World War II. The radiographic findings include pulmonary parenchymal changes and/or pleural changes.

The pulmonary parenchymal changes, which consist of small irregular opacities, begin in the basilar portions of the lungs; as the disease progresses, all portions of the lung may become involved. This is not uncommonly associated with a "honeycomb" appearance which consists of very fine small radiolucent areas characteristic of advanced interstitial fibrosis.

Optimum film quality is a prerequisite not only for detection of the early appearance of irregular opacities but also for detection of honeycombing of the lung.

The pleural changes include diffuse pleural thickening, localized pleural thickening in the form of plaques, pleural effusions, and mesothelioma formation. The early pleural changes are often very subtle; these minimal changes can easily be overlooked by the untrained physician, particularly if the costopleural junction is not well outlined on the radiograph. Pleural plaques may be seen in an "en face" projection or in a "profile" projection. The use of oblique views is important not only to confirm suspected pleural plaques on the PA projection but also to detect unsuspected plaques not visualized on the routine PA screening film. (Oblique views will also help confirm suspected areas of small irregular opacities not easily distinguishable from blood vessels on the PA view.) Characteristically, early pleural plaques occur in relatively asymptomatic patients, more commonly with no radiographic evidence of associated interstitial disease. In addition, noncalcified pleural plaques are more common than calcified plaques, although calcification makes them more readily detectable. Benign pleural effusions do occur some 10 - 20 years after asbestos dust exposure and in many instances are not associated with mesotheliomas. However, mesotheliomas do occur and, because mesotheliomas are most often the result of asbestos dust exposure, all pleural effusions should be suspect for mesothelioma formation until proven otherwise.

There is a high instance of bronchogenic carcinoma in patients exposed to asbestos dust, particularly if the patients are smokers. The finding of pleural plaques, either calcified or noncalcified, is merely a sign or a marker that the patient has been exposed to asbestos dust. While a higher incidence of bronchogenic carcinomas and mesotheliomas associated with pleural plaques has been reported, the plaque itself does not constitute the source of neoplasia. Patients with pleural plaques should have yearly routine chest films not only to record any progressive changes but, more important, to detect any early pulmonary carcinoma or mesothelioma.

LUNG CANCER SCREENING

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Lung cancer is the deadliest cancer in the United States. The American Cancer Society estimates that, in 1980, 117,000 new cases of lung cancer will be reported and 101,300 patients will die of this disease. Twenty-five percent of all cancer deaths are presently due to lung cancer; at least 80 percent of these could be prevented in the future by the cessation of smoking. However, there are indications that smoking is not decreasing and may be on the rise. Even if current programs against smoking are effective, the consequences of smoking will continue to be a major problem for a number of years.

Currently, the only hope for cure in these patients is early detection and prompt surgical removal. Radiotherapy, chemotherapy, and immunotherapy have not proven to be effective treatments for localized lung cancer.

It is generally accepted that symptomatic lung cancer is advanced cancer. In several series, the survival rate of patients with asymptomatic lung cancer is 3 to 4 times that of patients presenting with symptoms related to their cancer.

The size of the resected bronchogenic carcinoma is an important variable in overall survival. In Steele's study, if the resected bronchogenic carcinoma was 2 cm or less, the 5-year survival rate was 53 percent. Jackman reported a 68 percent 5-year survival in patients with resected bronchogenic carcinomas 2 cm or less in diameter, while the 5-year survival rate in Buell's study for cancers 2 cm or less in diameter approached 80 percent. It is evident then that the patients with the best prognosis are those with small asymptomatic lung cancers.

Since bronchogenic carcinoma may shed malignant cells into the sputum years before they become radiologically apparent, sputum cytology is also of importance in detecting these early lesions.

Screening of asymptomatic populations to detect early lung cancer has been attempted several times. Beginning in 1951, the Philadelphia Pulmonary Neoplasm Research Project (PPNRP) studied 6,136 males, 45-years of age or older, by semiannual photofluorograms of the chest for 10 years. The study concluded that radiographic screening of this type was not of value since only 6 percent of the 94 men who developed new histologically proven lung cancer during that decade survived 5 years. The PPNRP is frequently quoted as the *definitive study showing that radiographic screening is of no value in affecting the survival* rate in patients with lung cancer. However, on closer inspection, a number of inadequacies of this study are apparent:

1. The photofluorogram has been virtually abandoned in this country as a screening method.

2. Double reading of the photofluorograms was not performed during the first 5 years of the program.

3. There were no controls in this study.

4. The relationship of smoking to the development of lung cancer was not as well understood at the time of their study. Only 15.9 percent of the men smoked more than a pack of cigarettes per day.

5. Semiannual screening was not easily accomplished. Of the 94 patients that developed lung cancer, only half had a previous photofluorogram within 6 months, while a third had not had one within 12 months.

6. Unfortunately, of those 45 men who underwent surgery, 9 died within the first month after surgery, giving an unusually high operative mortality of 20 percent.

At the same time two less recognized and quoted, but more encouraging, radiographic studies were being conducted in London. In the North London Study, the study group had a 5-year survival of 23 percent, while the control group's 5-year survival was only 6 percent. In the South London Lung Cancer Study, the study group had a 4-year survival rate of 27 percent while the control group's was 6 percent.

In the late 1950's sputum cytology was becoming recognized as an accurate way of diagnosing lung cancer. In 1958, Lilienfeld et al. began studying 14,607 patients in 5 Veterans Administration domiciles with chest radiographs and sputum cytology every 6 months for 3 years. They concluded that sputum cytology could be applied as a mass screening method. Unfortunately, the techniques available at that time (rigid bronchoscopy and Dionosil bronchography) frequently could not localize the source of malignant cells when the chest radiograph did not give the exact location. In thirteen of 34 patients (38 percent) with a positive sputum cytology and a negative chest radiograph the source was not localized.

In 1968, Ikeda introduced the fiberoptic bronchoscope. In 1969, The Johns Hopkins Hospital with the aid of a National Cancer Institute grant began a pilot study to evaluate this instrument in the localization of radiologically occult lung cancer. We showed that this new instrument, and to a lesser extent bronchography with the experimental contrast agent tantalum, could successfully localize the source of the malignant cells in the patient with a positive cytology and a negative chest radiograph. The National Cancer Institute then issued contracts to three centers for the detection of early lung cancer: The Johns Hopkins Hospital (Johns Hopkins Lung Project - JHLP), Mayo Clinic (Mayo Lung Project - MLP), and Memorial Sloan-Kettering Hospital (National Lung Program - NLP).

This cooperative group is screening individuals who have a high risk of developing lung cancer (males 45 years of age or older who currently smoke a pack or more of cigarettes per day, or have quit smoking a pack or more per day during the year before entering the program).

At Johns Hopkins Hospital and Memorial Hospital the goal is to determine if the addition of frequent (every 4 months) sputum cytology to annual chest radiographs (considered standard procedure in many centers) will reduce the death rate from lung cancer when compared to the control group who only have annual chest x-rays. Through a variety of recruiting methods (motor vehicle records; contacting employers of large numbers of probable candidates, e.g., New York City Police Department, Baltimore Gas and Electric Company; and TV ads) each center has recruited over 10,000 eligible candidates. Before entering the program, they are randomized into a study and control group. After randomization, each candidate fills out a health questionnaire with the aid of a trained interviewer. In addition to identifying data, the questionnaire includes information about age, race, current symptoms of cough and sputum production, smoking habits, occupational exposures and a previous history of chest illness or lung cancer. An individual was disqualified if age or smoking habits did not comply with our criteria or there was a history of a previous lung cancer.

The study group has an induced sputum cytology (aerosol of Hank's solution) and then is instructed to collect 3 days of sputum at home in a bottle of Saccomanno's fixative. This is returned to the project in a prepaid mailer. The study group receives a PA and lateral chest radiograph, generally done at 140 kVp. The control group has only a chest radiograph. The study group is mailed containers to collect 3-day pooled sputum cytology at 4 and 8 months. At 12 months the screenees return to the hospital and again are screened with induced sputum cytology, a 3-day pooled sputum, health questionnaires, and a PA and lateral chest radiograph. This cycle will be continued for 5 years. The control group will be followed only by annual PA and lateral chest radiographs and health questionnaires.

In the JHLP, all of the radiographs are independently double read by two members of a pool of board certified diagnostic radiologists. If any abnormalities suggestive of or suspicious of lung cancer are detected, the films are sent to an independent referee for final disposition.

In the NLP, the radiographs are obtained by a specially trained technologist who then evaluates them. On the following day the radiographs are reviewed by a board certified diagnostic radiologist at Memorial Hospital. Those radiographs which contain an abnormality, or about which there is any difference of opinion between technologist and reading radiologist are referred to the NLP study radiologist for final disposition.

At the Mayo Clinic the aim is to determine, if intensive cytologic and radiologic screening can significantly lower the death rate from lung cancer. The MLP interviewed 11,122 "high risk" patients entering the Mayo Clinic. Of these, 9,313 were included into the project. Those excluded were primarily referred to the Mayo Clinic with the diagnosis of lung cancer, or a life expectancy of less than 5 years, or those that would be unable to tolerate a pulmonary resection. The remaining 9,313 were randomized into a study and control group. Initially the study and control groups were screened identically. Each had a chest radiograph, 3-day pooled sputum cytology, and a health questionnaire. After the initial screening, the study group is requested to submit a PA and lateral chest radiograph and a 3-day pooled sputum cytology, as well as complete a health questionnaire every 4 months. The control group is only advised to have an annual chest radiograph and sputum cytology, and is followed by means of a questionnaire alone. This program is also to continue for 5 years.

All of the initial radiographs are done at the Mayo Clinic. Seventy-five percent of the follow-up radiographs are done in areas remote from the Mayo Clinic and are paid for by the MLP and mailed to the Mayo Clinic. The radiographs are independently double read by two members of a pool of radiologists, experienced pulmonary internists, and a trained technologist. Any abnormalities suggestive of or suspicious of cancer are sent to another member of the pool for final disposition.

At all three centers conditions known or thought to predispose to lung cancer, or be associated with lung cancer, are also recorded (e.g., asbestosis, healed inflammation, chronic obstructive pulmonary disease).

All three centers have now completed their intitial recruitment (prevalence). In most hospital based series, the resectability rate of prevalence cases for patients with bronchogenic carcinoma is 33 percent with rates as low as 12 percent having been reported. The resectability rate in each center is about 60 percent, a two-to-five fold increase in resectability. The incidence portion of the project (new cases of lung cancer developing while in the project) also shows at least a doubling of the resectability rate of the study groups over the general population.

The early results from the three-center NCI cooperative group to detect early lung cancer are encouraging, but must be interpreted with caution because they are based on resectability rates rather than survival rates or mortality rates. Several tentative conclusions can be made at this time but are not to be considered as recommendations.

1. Screening should include both radiography and sputum cytology because the two techniques are complementary. Sputum cytology is useful in detecting central lesions, which are usually squamous carcinomas; chest radiographs detect peripheral lesions, which are usually adenocarcinomas. There is very little overlap between these two screening techniques; however, chest radiography detects 75-80 percent of all lung cancers.

2. The radiographic appearance of early lung cancer is protean. Any radiodensity in the lungs of the high-risk patient (male or female), especially if it is a new lesion, should be considered lung cancer until proven otherwise.

3. The lateral chest radiograph is important in this age group.

4. Double reading of the chest radiographs has been very helpful.

5. Currently, no definite conclusions as to frequency of screening can be made. It is hoped that in the future when the data obtained from the questionnaires (age, smoking habits, and industrial exposure) are correlated with the radiographic appearance of the chest (e.g., old inflammation, asbestosis) and the degree of atypia of the sputum cytology, specific recommendations will be made for certain subclasses of patients.

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DISCUSSION

DR. KELSEY: Can you tell us something about your retake rate in this lung cancer survey?

DR. STITIK: The initial prevalence screening, I am afraid, was done in two centers. One center employed primarily 120 kVp, with an unsatisfactory rate of about 11 percent. The other portion of the screening was done at Johns Hopkins Hospital; here we employed 140 kVp and the rate was less than 1 percent. We now do all our films at Hopkins and the retake rate is less than 1 percent. Many retakes occurred at the first screening when we would cut off a costophrenic angle in a devoted chest room. When these volunteers return they say, "You have to do me sideways."

DR. GREGG: Considering the growth rate of these tumors, how often do you advise chest radiographs?

DR. STITIK: This is a question that will probably be better answered at the Mayo Clinic where they take more frequent x rays. It is my impression that there are some lung cancers (e.g., oat cell, small cell) that you are not going to cure even if you take films frequently. There are, however, other lesions that change very slowly, usually adenocarcinomas. I think, when everything is finally evaluated, that examination every 6 months will be optimal. You are probably not going to help patients with oat cell carcinoma, but at 6 months a patient with adenocarcinoma may be helped.

You run the risk, of course, of aggravating patients by bringing them back too frequently, but 6 months seems reasonable. Sputum cytology should obviously be done in addition to, rather than in place of, a chest x ray as the two procedures are complementary.

MR. SHAVER: I have a question for Dr. Sargent. If you were the attending physician of a 29 year old who showed asbestosis, how often would you recommend a radiograph?

DR. SARGENT: This is my own personal opinion, but I would say about once a year.

MR. SHAVER: For what purpose?

DR. SARGENT: To document progression of disease and detect possible neoplasia. There is no treatment for asbestosis. The only thing you can do is tell them to stop smoking.

DR. ALTER: Would you not advise that for anyone?

DR. SARGENT: To stop smoking? Of course.

DR. STITIK: Dr. Muhm, do you have any better answer to Dr. Gregg's question?

DR. MUHM: I would agree with what you say, Dr. Stitik, because we find at the Mayo Clinic that some subjects have an absolutely negative chest x ray and then, 4 months later, there is a 4-cm, small cell carcinoma of the hilum. In other cases, we find cancers that have been present on the chest x ray for 20 months, and we missed them in spite of taking an x ray every 4 months. We missed them with three physicians looking at each of five chest x rays taken at 4-month intervals. I think that every 4 months is probably too often and every year is probably not quite enough. So, every 6 to 8 months is what we come up with.

DR. RAO: Does any physician conduct a physical examination on these patients?

DR. STITIK: No.

DR. MILNE: I must have missed something in how you selected your control series. You seem to have as many cancers in the control series as in the noncontrolled, yet I thought when you started off that the controls must be non-smokers. Am I wrong?

DR. STITIK: Yes. As I indicated, the subjects are randomized before they come into the study. Someone responds to our television ads, our newspaper ads, or mass mailings, and wants to join our project. We then make sure he smokes, is over the age of 45, and does not have a previous lung cancer. If he fits our criteria, he is randomized into a study or control group before we even see him.

The University of Cincinnati has looked at our randomization and it really could not be any better. We have the same number of 50 year olds, the same number of 1 to 1.5 pack-per-day smokers, and so forth, in each group. It was an excellent randomization. They all smoke; having 34 lung cancers in the control group and 33 in the study group does not upset me.

DR. MILNE: What is controlled about the fact that one group gets a full set of examinations and the other does not?

DR. STITIK: The control group does not have sputum cytology, but they all have annual chest x rays.

DR. ALTER: That was a prevalence study, is that right? You do not have the data on your incidence yet, do you?

DR. STITIK: Our early incidence data are approximately the same as the prevalence data. We are continuing to operate on 60 percent, and 75 to 80 percent of all the bronchogenic carcinomas are detected by chest x rays.

DR. ALTER: Do you think if you had a less selected population, including some nonsmokers, the costs of detection and cure would be higher?

DR. STITIK: Sure!

DR. SARGENT: Cost must be considered in taking chest x rays of asbestos workers, whether once a year or every 6 months. What difference does it make which interval is used? Have you any data to suggest that an x ray every 6 months makes much difference?

DR. STITIK: We actually wait 12 months, which concerns me. When we first designed the program, I favored having fairly frequent exams, that is, radiographs every 4 months at Hopkins as they are done at the Mayo Clinic. The problem with the Mayo Clinic arm of this project is that very few of their volunteers live in Rochester, Minnesota, so the Clinic has to rely on mailed-in films. Initially, some of the mailed-in films were atrocious. I had wanted to have this frequent radiographic screening done at Hopkins, where we would use 140-kVp devoted chest x-ray units. Dr. Muhm and I both suggested 6 months, yet in practice we are waiting a full 12 months at Hopkins and Memorial.

DR. RAVIN: Dr. Muhm's presentation suggested that analysis of the margins of lesions could be quite helpful in establishing the diagnosis of malignant or benign. Based on your experience with screening early cancers, could you comment on the reliability of the tumor/lung interface as a distinguishing feature?

DR. STITIK: As most of you know, it is my personal opinion that in about 85 percent of cases the radiologist can tell lung cancer from benign disease, but such accuracy is not good enough. This is why we resort to needle biopsy, bronchoscopy or thoracotomy to decide what a lesion is, rather than depend upon analysis of its interface with normal lung.

DR. NELSON: You have both alluded to putting the 14 x 17 cassette sideways for obliques and large people. Loren Niklason's work indicates that, when you put the grid strips perpendicular to the mediastinum, cleanup of scatter is less effective there. Yet, scatter in the diaphragm, which would then be parallel to the strips, would be cut by about 50 percent. I wonder if you are picking up that diaphragmatic calcification merely by positioning, because of this effect. Because of the scatter phenomenon, I wonder whether we should have a 14 x 17 chest cassette just for big people, with "horizontal" grid strips. It would make quite a difference. What was that figure?

MR. NIKLASON: Grid orientation may affect the amount of scattered radiation under the mediastinum by about 25 percent.

DR. SARGENT: I think we are seeing calcification mainly on oblique views because the beam is tangential to it. I do not think that scatter effect makes a difference. The other problem is that sometimes you take two films because the patient's chest is too wide.

DR. GORDON JOHNSON: I thought I heard you recommend obliques. What are the criteria, or when do you recommend them?

DR. SARGENT: I do not have the answer to that. We rarely do a study with only one view. I feel that in certain cases we pick up noncalcified plaques that cannot be seen on a PA film. I do not know what percentage of cases fall into this category. As a routine screening procedure, we recommend a single PA film; if there is any suspicious lesion, an oblique should then be done. On the other hand, in some cases lesions are identified only on the oblique.

CHARACTERISTICS OF SCREEN/FILM SYSTEMS FOR CHEST RADIOGRAPHY

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A film representative once approached a radiology administrator and said, "Mr. Green, our company has just come up with a couple of new films which I would like you to evaluate for us."

"Certainly," said Mr. Green. "I'll be glad to arrange the tests for you. As a matter of fact, my secretary has just recovered from a case of pneumonia and I'll ask her if she would like to be a volunteer for the study." The secretary readily agreed and the films were made. The film representative showed the two films to a chest radiologist and asked him which one of the films he liked better.

"This patient still has a trace of pneumonia in the left lung and the right lung is completely clear. I can see the presence of pneumonia in the left lung much better in Film A than in Film B," the chest radiologist replied. Thereupon, the film representative took the same two films to a bone radiologist. The bone radiologist looked at them and said "Look, there is a hairline fracture in one of the ribs here. It is clearly demonstrated in Film B but not in Film A. I would certainly vote for Film B."

Meanwhile, a radiation physicist happened to pass by and the film representative requested him to evaluate the films as well. The physicist took the films back to his laboratory, kept them there for a couple of weeks, performed a series of measurements and finally concluded that the information content on the films was identical.

Finally, the representative took the films to the radiology administrator. The administrator flipped open the envelope, placed the films on a view box, quickly turned around and said "John, neither one of these films is any good. I happen to know that this young lady just broke up with her boyfriend the day before you made the films and neither film shows a broken heart." The moral of the story is that what is good for the goose is not good for the gander. Chest radiologists differ widely in their preferences of technique and image receptor.

In objective terms, however, five basic parameters that govern image quality can be identified: density, contrast, latitude, resolution and noise. These parameters, as visualized on the finished product, depend on a complex interplay of the generator controls selected, geometric factors, focal-spot size, methods used for scatter elimination and image receptor characteristics. There is also considerable interplay among the image parameters themselves, a fact that is frequently not understood.

FILM DENSITY

Adequate film density is by far the most important of the five basic characteristics. Most repeat studies in radiology are due to improper positioning, patient motion or inadequate density. If the positioning is correct, the film density is adequate, there is no patient motion or tube vibration and there are no visible processing defects, the film is accepted. Considerations of latitude, contrast, resolution and noise come into the picture only during the initial setting up of the routines and the technique charts. Most radiologists agree that, for a given radiograph, the average film density should be about 1.0, with a range of about 0.25 to 2.0. Adequate average density can be obtained by manipulating a set of variables that numbers up to 13, depending on thickness of the body part, patient's age and pathology. The variables are: generator type (single-phase or three phase), tube potential, mA, time, beam filtration, focus-to-film distance, object-to-film distance, field size, grid, developing time, developing temperature, developer chemistry and image receptor. In practice, with a given piece of equipment and a given procedure, the kVp, mA, and time are the only variables the technologists manipulate to produce a film with adequate density. Within a given institution, the image receptor is also generally the same. With large patients, a grid is usually interposed between subject and image receptor.

CONTRAST AND LATITUDE

Contrast and latitude go hand in hand. The contrast in any given region of a radiograph may be defined as the difference between optical densities within a given anatomical structure located in that region, and the immediate surroundings. Latitude on the other hand is a measure of the range of anatomical detail that can be simultaneously visualized. A high-contrast image usually has poor latitude and vice versa. These parameters, as visualized on the radiograph, depend on the image receptor used and the tube potential selected. As far as the image receptor is concerned, it is known that contrast and latitude are governed only by film and are independent of the screens and beam quality, provided that film darkening due to direct x-ray absorption in the film is negligible compared to that due to light emission from the screens (1). Film latitude and contrast are subject to considerable variation depending on development conditions. In an actual radiographic situation, the range of tissue densities displayed in different regions of the radiograph and the contrast variations among them are determined by the film and the tube potential. With a film exposed at low tube potential, the contrast is higher in certain regions of the radiograph but the overall latitude is lower. At high tube potential, the latitude is higher and the contrast lower. Subtle differences in contrast and latitude cannot be appreciated unless the radiographs in question have identical optical densities at a given reference point. Furthermore, in an underexposed or overexposed radiograph, variations in tissue density are not displayed as proportional variations in film density because of the "toe" and "shoulder" regions of the characteristic curve. For these reasons, it is meaningless to discuss latitude and contrast unless the radiographs in question are optimally exposed with respect to optical density:

In chest radiology, image contrast and image latitude influence diagnostic accuracy in the following manner.

A high-contrast chest radiograph will tend to overexpose the lungs and underexpose the mediastinum, both of which can lead to missed diagnoses. A high-latitude film of proper density will allow interpretation of all structures seen on the chest radiograph, but will make it more difficult to analyze subtle changes within the mediastinum and soft tissues. In clinical practice, a high-latitude film is much less objectionable than a high-contrast film.

Because contrast and latitude can be manipulated significantly by altering the tube potential, there has been considerable discussion in the literature on this subject. The standard technique for PA chest radiography from its inception was 70-80 kVp and no grid. In recent years, many radiologists have switched to 110-130 kVp techniques with a 12:1 grid. In 1970, in a paper presented on behalf of a subcommittee of the Union Internationale Centre Le Cancer Committee on Asbestos and Cancer, members of the McGill University Asbestos Study and the Roentgenologic Panel of the United States Public Health Service Asbestos Study, Jacobson et al. compared 60-80 kVp techniques with 120-145 kVp techniques (2). They found that:

> "The low kilovoltage technique has several disadvantages. First, it does not permit the use of short exposure times to avoid motion, e.g., less than 1/30 of a second, except when generators of considerably higher capacity than 300 mA are used or when the patient is quite thin. Second, radiation increases rapidly with the larger patients or with use of more than 75 kVp, thus decreasing radiographic detail. The reduction of radiographically effective radiation resulting from the interposition of a grid or air gap

> between the patient and the film to eliminate this secondary radiation must be compensated for by either increasing the exposure time which is already relatively long or increasing the kilovoltage. The high kilovoltage technique on the other hand decreases the effects of technical errors, such as selection of wrong exposure factors, dropping line voltage, poorly functioning phototimer, etc. For these reasons, the technique used should combine as high a range of kilovoltage and as low a range of milliampere-seconds (mAs) as permitted by available equipment. This produces radiographs with maximum detail and minimizes variations in density and contrast, so that radiographs are consistent in quality and the number of unsatisfactory examinations is reduced. The latter is probably the most outstanding feature of the high kilovoltage technique."

In 1977, Christensen et al. made a clinical comparison of the 70 kVp technique without a grid and a 120 kVp technique with a grid. Using a phantom and beeswax nodules ranging from 3-7 mm, they found that "more nodules of all sizes except 3 mm were detected with the 120 kVp technique. The 3 mm nodules were rarely detected with either technique" (3). On the other hand, almost twice as many false positive nodules were detected with the 120 kVp technique as with the 70 kVp technique. The radiation dose was also 50 percent higher, but this may be partially offset by a reduction in repeat films.

More recently, there have been several attempts to use the new 350 kV field-emission x-ray tubes in chest radiology. Commenting on the use of the 350 kV techniques, Hallenbeck writes:

> "The 350 kVp units permit improved resolution of pulmonary vessels, facilitating earlier and more accurate diagnosis of congestive heart failure and other cardiac conditions. Diagnosis of small lesions was improved, particularly those within bone shadows. Clear cut separation of vascular and non-vascular shadows in the hilar region, where lesions of considerable size can sometimes be difficult to discriminate, permitted rapid detection of hilar adenopathy and bronchogenic neoplasms. Using the new system, the percentage of retakes has been cut to 2-3 percent and most of these are necessitated by the patient being too large for the chest to be fully encompassed by the vertical film configuration employed with the DuPont changer. Fluctuation in the phototimer setting is insignificant, resulting in excellent reproducibility of radiographic technique over the course of several examinations. This makes it much easier to compare radiographic findings than was true with the earlier system, in which there were wide ranges in radiographic technique despite a vigilant quality-control program. The 350 kV units have given consistently good quality and properly exposed radiographs in patients of all ages" (4). He admits, however, that:

> "There was a tendency to overread the 350 kV films due largely to the greatly increased amount of information. The result was heightened suspicion of miliary infiltrates in patients who were actually normal, while the improved visualization of the lung apices led to a general suspicion of increased incidence of disease in these areas which was not verified by other examinations" (4). However, he claims that:

> "After about four to six weeks of experience, the 350 kV radiographs could be evaluated with no more difficulty or time than was required for the low kV films. Calcifications within small lesions have usually had to be verified by obtaining either a low kVp spot film or a tomogram of the area. Through experience, however, calcifications can often be readily detected on the 350 kV chest films. Radiologists specializing in the study of bone or the gastrointestinal tract required more experience to adjust to the apparent lack of contrast and the overall gray tone of the 350 kV chest films; however, this too was quickly resolved by constant practice" (4).

On the basis of several clinical studies, Proto and Lane conclude that the three main disadvantages of the 350 kV technique are: (1) slight mottle, usually recognized in soft tissue areas of thorax; (2) reduced visualization of calcium, which improved with experience (some overall loss was found, as well as an overall gain in areas that are underpenetrated in routine studies); and (3) general loss of rib detail, but better visualization of spine in areas routinely underpenetrated (5). Advantages include: (1) better visualization of nodules, vascularity, interstitial and alveolar disease, mediastinal reflections, infradiaphragmatic and retrocardiac areas, larynx and trachea, and pleural and cardiac abnormalities; (2) reduction in retakes; (3) facilitation of patient flow; (4) simplicity of installation and operation; (5) uniformity between examinations; and (6) decrease in patient exposure.

Studies by Christensen et al., however, did not reveal significant clinical advantages (6). "Nodule detectability improved with increasing energies up to 200 kVp, but the improvement was relatively small and was accompanied by an almost equal rise in the number of false positive readings. Patient exposures were least at 200 kVp. There seems to be no advantage in a 300 kVp technique, because nodule detectability decreased and patient exposures increased at this energy."

In summary, there is by no means universal agreement as to which is the best kVp for chest radiographs. As stated by Proto and Lane, no single technique seems to give optimal visualization of all areas in a chest radiograph (5).

IMAGE RESOLUTION

In a chest radiograph, a sharp line of demarcation, e.g., a rib margin, often appears as a diffuse border. Furthermore, if the anatomical structures of interest are small and are positioned close to each other as in the case of miliary lung densities or the bone trabeculae, they may fuse into each other in the image. For this reason, image resolution is important in chest radiology with reference to the detection of hairline fractures and metabolic bone disease; in miliary, fibronodular or interstitial lung disease; and in vascular definition in early pulmonary venous hypertension.

The terms "unsharpness," "resolving power," "line pair resolution" and the "modulation transfer function" are used to describe these features of the image. Among these, the modulation transfer function is the most recent and most useful.

The modulation transfer function (MTF) of an imaging system can be defined as a graph that describes the percent contrast that is retained as test objects of sinusoidal transmission and progressively increasing spatial frequency (cycles/millimeter) are imaged. In clinical radiography, it depends on a complex interplay of various factors, such as focal spot size and shape, magnification factor, intensifying screens used and the degree of patient motion involved. The MTF of an x-ray film/screen combination depends primarily on the type of screen used and to an insignificant degree on the type of film. This is because the resolution capabilities of x-ray film are far superior to those of intensifying screens. The MTF of a radiographic screen is determined by the degree of light diffusion that occurs between the screen and the film. This, in turn, is governed by the chemical composition of the screen material, size of the crystals, screen thickness, the tightness of the packing and the presence of any dye. Poor film/screen contact is a common cause of MTF degradation, making excellent film/screen contact important in clinical radiography. Even though the MTF of a film/screen combination is primarily determined by the screens used and is independent of the film, image sharpness as visualized on a clinical radiograph does depend on the film. This is because the MTF is normalized to unity at zero spatial frequency and hence does not take into account the fact that image sharpness depends also on the image contrast at the point of observation. The optical density at the point of observation also affects the visualization of image sharpness. For these reasons, subtle differences in image receptors with respect to resolution cannot be appreciated unless the radiographs in question have identical optical densities and contrast at a given reference point. In other words, if one wishes to compare two intensifying screens, they must be used with the same patient or phantom, same film, same geometry, same focal spot and same kVp. Only the mAs should be varied, again making sure that the time settings involved are not large enough to produce appreciable motion blurring.

Earlier it was shown that the film should be selected on the basis of tube potential preference. As far as selection of the intensifying screens is concerned, it must be remembered that this ties in very closely with the selection of the focal spot because the loss of resolution due to the image receptor depends on the nature of the intensifying screens used. In general, the faster the screen, the poorer the resolution. However, the faster the screens, the smaller is the exposure required and the less the loss of resolution due to patient motion. Patient motion in turn depends on the velocity of motion, the type of motion and the exposure time. In addition to the intensifying screens, the exposure time depends on the focal-spot size as well. The larger the focal-spot size, the larger the maximum tube current at which it can be used and the smaller the exposure time. On the other hand, the larger the focal-spot size, the larger is the loss of resolution due to geometric factors. The loss of resolution due to geometric factors depends also on the magnification involved in the imaging process. At a larger magnification factor, the resolution is poorer. On the other hand, the larger the magnification factor, the larger is the image that is presented to the image receptor and the better the utilization of its resolution capability. For these various reasons, it is important that a careful analysis be made before one decides on the optimum parameters to be used in a given clinical radiographic situation. In the case of chest radiography, there is very little magnification and so the focal spot size is not very critical. The only parameter that needs to be considered is the trade-off between intensifyng screens and exposure time. With faster intensifying screens, poorer resolution, lower exposure time and less motion unsharpness result. At extremely low exposure times, however, exposure reproducibility becomes a problem, especially when phototimers are used.

RADIOGRAPHIC NOISE

The noisy appearance (radiographic mottle) of a uniformly exposed film sandwiched between a pair of intensifying screens is a result of: (1) quantum mottle which arises due to the random spatial fluctuations in the number of x-ray photons; (2) screen mottle which is due to structural inhomogeneities in the screen phosphor material; and (3) film graininess due to the random distribution of exposed grains in the film emulsion. Quality control in the manufacturing of screens is generally good, so the mottle due to structural inhomogeneties in the screen phosphor is minimal. The spatial variation of the density on a film which was exposed in contact with a pair of intensifying screens, therefore, consists of a coarse pattern caused by quantum mottle superimposed on the fine grain pattern of the film. A number of studies by Cleare, Rossmann and others have shown that the quantum mottle depends not only on the number of x-ray and light quanta involved but also on the MTF of the screens (7, 8, 9).

The overall mottle in an exposed film can be measured in terms of the Selwyn noise index. This is defined as the quantity $A\sigma_D^2$ where σ_D is the standard deviation of a number of density measurements on the film and A is the area of the aperture used in making the measurements. This is a gross index and does not reveal the spatial frequency content of the noise and its dependence on the MTF of the screens. The random distribution of the x-ray photons absorbed in the screens corresponds to a "white noise" whose manifestation on the film depends on the MTF of the screens and the gradient of the film's characteristic curve. For this reason, a more sophisticated method, involving the conversion of density fluctuations on a uniformly exposed film into their Fourier components, is often used to characterize the noise in it. The square of the Fourier transform of the measured density fluctuations is referred to as the Wiener spectrum, which provides a complete description of the spectral composition of noise. Wagner et al. (10) and Doi et al. (11) have recently measured the Wiener spectra for a number of modern x-ray film/screen combinations.

In the case of chest radiography, presence of quantum mottle can interfere primarily with the recognition of bone and soft tissue details. The bone appears "washed out" in a pattern suggesting demineralization as in osteomalacia. Similarly, mottle can interfere with the visualization of interstitial patterns within the lung and the soft tissues. However, the mottle can usually be compensated for by the radiologist, and it is relatively easy to recognize this potential error because the mottle is visible over the entire film.

INTER-RELATIONSHIPS BETWEEN QUANTUM MOTTLE, SPEED, RESOLUTION AND SCREEN ABSORPTION

Although the resolution characteristics of x-ray intensifyng screens are best specified by the complete MTF curve, in practice it is adequate to quote only one or two points on the curve. For example, the spatial frequency (cycles/mm) at which the MTF becomes 0.5 and 0.1 may be quoted. These are referred to as f_{50} and f_{10}, respectively. Another single number measure of image sharpness is the equivalent passband N_e, defined as the integral of the squared MTF:

$$N_e = \int_0^\infty (M(f))^2 \, df$$

f_{50} and f_{10} bear a constant relationship to N_e in case of exponential approximations of the MTF, i.e., $f_{50} = 0.693N_e$ and $f_{10} = 2.3N_e$.

Recently, Rao et al. established that the equivalent passband of a pair of intensifying screens is inversely related to its speed (12). More specifically, they found that the equivalent passband of a pair of intensifying screens of a given phosphor material is inversely proportional to the square root of the system speed when used with a given film, i.e., $N_e\sqrt{S}$ = constant. Straight lines passing through the origin are obtained when one joins the points for different screens of the same phosphor material used with a given film. The slope of the straight line for a given phosphor material depends on the film used and the quality of the x-ray beam incident on the image receptor.

The integral of the Wiener Spectrum above the film granularity level yields the noise variance. Rao et al. recently demonstrated that the root mean square noise (i.e., the square root of the variance) is proportional to $\gamma N_e \sqrt{S/\varepsilon}$ where ε is the screen absorption factor, S the speed of the screen/film combination, and γ the slope of the characteristic curve at the point of interest (13).

Even though the quantity $\gamma N_e\sqrt{S/\varepsilon}$ is an accurate measure of the mottle as would be observed on a uniformly exposed film, in reality the quantity of significance is $N_e\sqrt{S/\varepsilon}$. This is because the signals in the radiologic image and the noise are equally influenced by the film gamma and hence the signal/noise ratio is independent of it. For this reason, they have named this quantity the Quantum Mottle Index (QMI). Tables containing equivalent passbands, QMI values and speeds for various screen/film systems can be obtained from a 1978 article by Rao et al. (14).

With a given pair of screens, N_e and ε are constant regardless of the film, so that the QMI is directly proportional to the square root of the film. With a given film, the QMI depends on the values of $N_e \sqrt{S/\varepsilon}$ for the screens. However, if the screens are all made of the same phosphor material, the QMI can be seen to be proportional to $1/\sqrt{\varepsilon}$ because $N_e\sqrt{S}$ has been shown to be a constant in this case.

More recently, we have also been able to derive a quantitative relationship between speed and the screen absorption factor. More specifically, we found that the speed S of a screen/film system is approximately proportional to the screen absorption factor up to ε values varying from 0.4 to 0.6. Then speed decreases with ε because of increased light absorption within the screen. This relationship holds provided the same type of film is used with screens of a given phosphor material but of varying thickness. With the assumption that S is proportional to ε, it can be concluded that QMI is proportional to $1/\sqrt{S}$ for screens of a given phosphor material and a given film. On the other hand, with a given pair of screens, the QMI will be directly proportional to the square root of the film speed.

RADIATION DOSE

The ultimate object of this conference is to explore ways and means to reduce patient exposure in chest radiography without sacrificing diagnostic accuracy. The entrance exposure for an average postero-anterior view using DuPont HI-plus screens and medium speed film is about 25 millirads with 70 kVp/no grid/ 2 mm Al HVL, 40 millirads at 120 kVp /12:1 grid/3 mm Al HVL and 8 millirads with 350 kV/ 12:1 grid/11.0 mm HVL. Notice that the highest exposure occurs with the most commonly used technique. Use of faster screens is one way to reduce these exposures. Loss of resolution and poor phototimer reproducibility are among the undesirable consequences of such an approach. Quantum mottle and motion unsharpness, however, decrease with screen speed (see previous section). Keeping the screens fixed and using a faster film is another way to reduce doses with any technique. Increased quantum mottle and poor phototimer reproducibility are the undesirable consequences of this approach. Patient exposures can be reduced to extremely small values if one is prepared to use the fastest screens and the fastest film available on the market. Most radiologists find the attendant loss of resolution, increased quantum mottle, and poor phototimer reproducibility unacceptable. There is some justification for this. For example, using a thorax and lung phantom and simulated tissue equivalent 6.4 mm lesions, Gray et al. found that detection accuracy decreases and false positive rates and intraobserver disagreements increase with increased quantum mottle (15). On the other hand, Kundel et al. have shown that only 25 percent of all false negatives are attributable to pattern recognition errors, the remaining 75 percent being due to scanning and decisionmaking errors (16). It is, therefore, not clear to what extent diagnostic accuracy will truly be impaired by use of low-dose techniques. Extensive clinical trials alone can provide some answers to this question. Even with extensive clinical trials, we will perhaps only be able to answer the question in broad statistical terms, considering the complexity of the diagnostic process.

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THE RELATIONSHIP BETWEEN SCREEN/FILM SYSTEM AND GEOMETRIC UNSHARPNESS: CONVENTIONAL AND 350 kVp CHEST RADIOGRAPHY

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In chest radiography many physical factors have a significant effect on image unsharpness and patient radiation exposure. Image unsharpness is caused by recording system unsharpness, geometric unsharpness, and motion unsharpness. The selection of the screen/film system is important not only for image quality but also for considerations of patient radiation exposure, patient motion, x-ray tube output, and x-ray tube life. The effect of geometric unsharpness can play an important role in the selection of screen/film systems (1, 2). Geometric unsharpness (U_g) is determined by the size and shape of the intensity distribution of the x-ray tube focal spot in combination with the object-to-recording and focal spot-to-recording system distances.

We have evaluated the effects of screen/film and geometric unsharpness for two different types of x-ray units used for chest radiography in our department. A General Electric (G.E.) Maxiray 100 x-ray tube with nominal focal spots of 1.0 mm and 2.0 mm is used for chest radiography at 120 kVp. DuPont Hi Plus screens with DuPont 2 DC film are used with this technique. A Hewlett Packard (H.P.) Field Emission x-ray unit at 350 kVp is also used for chest radiography in our department (3). DuPont SP screens with DuPont 6 Plus film are used for this technique. The characteristic (H & D) curves displaying the relative speeds of the recording systems are shown in Figure 1.

Line spread functions (LSF's) of the focal spots were measured using the slit technique. For the 120 kVp G.E. focal spots, 1.6 mm thick Pb slit jaws were used with a 50 μm slit opening. For the 350 kVp H.P. unit, specially designed 7.7 mm thick tungsten jaws were used with a slit opening of 200 μm. The modulation transfer functions (MTF's) of the focal spots were calculated from the LSF's. The MTF's of U_g in the image plane were calculated from the MTF's of the focal spots and from the geometric magnification factors corresponding to several object-to-recording system distances or chest planes (Fig. 2). Pinhole pictures of the focal spots showed that the H.P. focal spot was doughnut-shaped and nearly symmetric. The G.E. unit focal spots displayed the usual double-band distribution. Results for the largest dimension, which is perpendicular to the cathode-to-anode axis, are shown in Figures 3 and 4 for the G.E. focal spots.

Figure 3 shows MTF's of U_g at midplane of the chest for the G.E. unit, large and small focal spots, and for the H.P. unit. Also shown are the MTF's of the screen/film systems. The Hi Plus and SP screen/film system MTF's were nearly identical. Also plotted is the MTF for the sharper Par speed screen system. The MTF of U_g for the H.P. unit is more limiting in resolution than both G.E. focal spots and both screen/film systems. The large G.E. focal spot and the Hi Plus screen/film MTF's are similar at the 10 percent level. The MTF's of the small G.E. focal spot and Par speed screen/film system have comparable resolution at the 10 percent level although the small focal spot MTF is better overall. The implications from these curves are that U_g is a significantly limiting resolution factor compared to certain screen/film systems especially for the Hewlett Packard unit. One should keep in mind that the MTF of U_g gets progressively worse as the object-to-recording system distance increases.

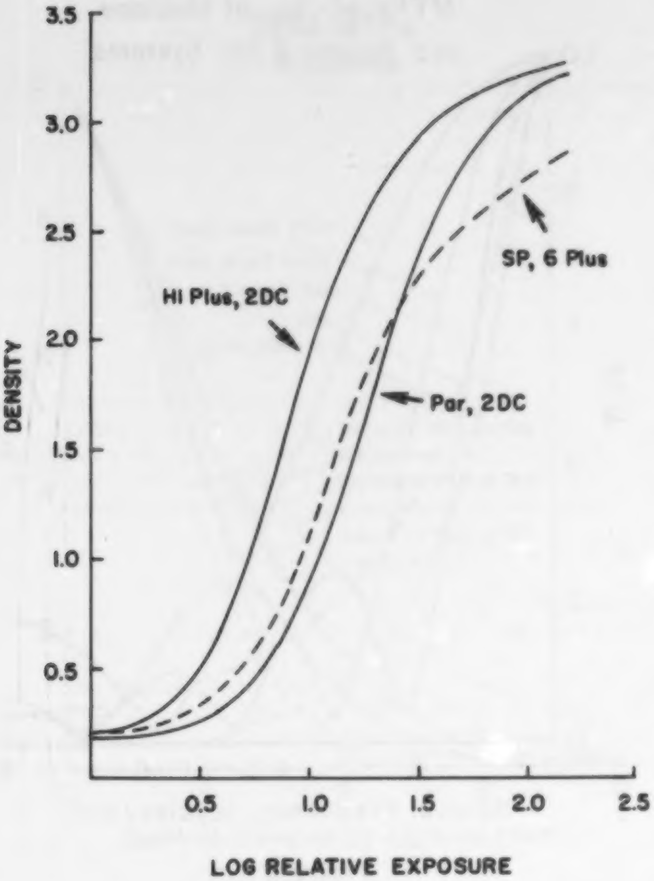

Figure 1

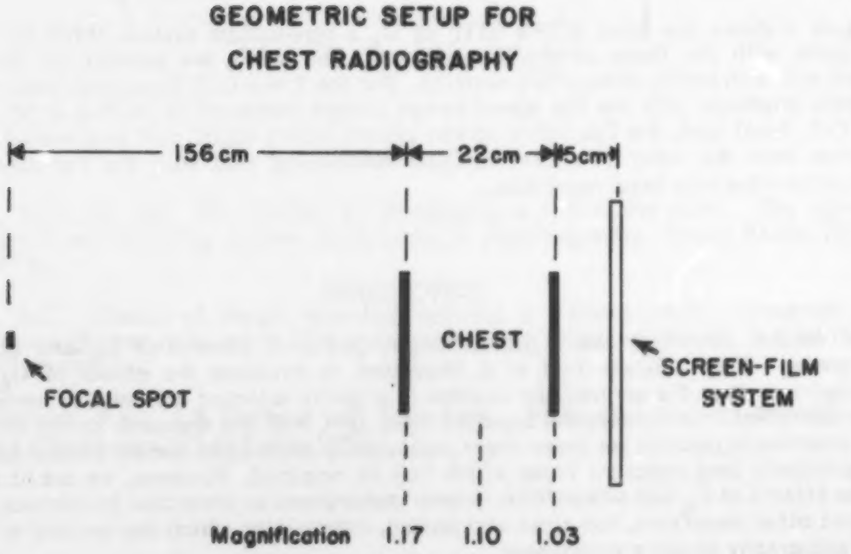
GEOMETRIC SETUP FOR
CHEST RADIOGRAPHY
156 cm
22 cm
5 cm
FOCAL SPOT
CHEST
SCREEN-FILM
SYSTEM
Magnification
1.17
1.10
1.03

Figure 2

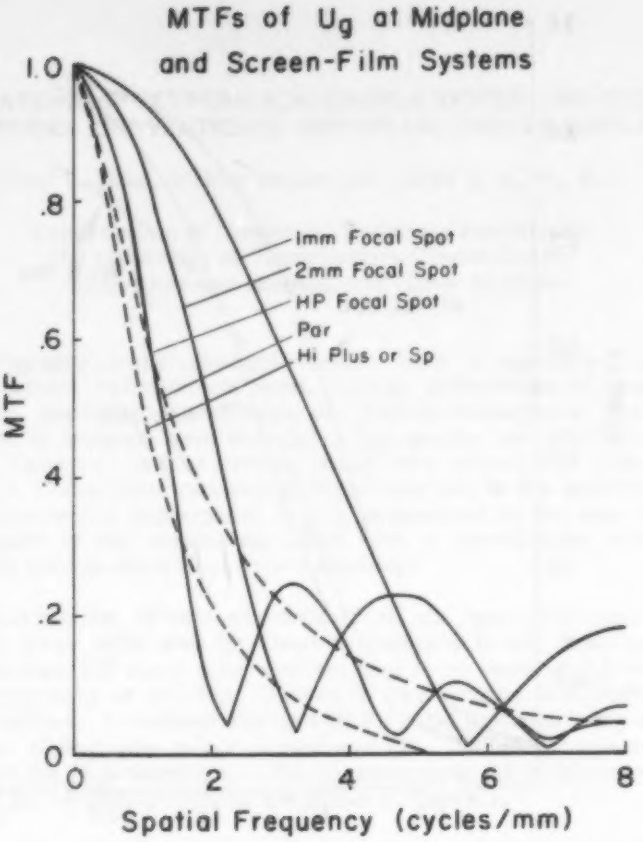

Figure 3

Figure 4 shows the total MTF's (MTF of U_g x screen/film system MTF) for the three focal spots with the three screen/film system. The MTF's are poorest for the Hewlett Packard unit with either screen/film systems. For the 2 mm G.E. focal spot total resolution is slightly improved with the Par speed screen system compared to Hi Plus or SP. With the 1 mm G.E. focal spot, the Par speed screen system offers significant improvement in total resolution over the other two screens. The small focal spot with the Par speed system clearly offers the best total resolution.

CONCLUSION

The results described above illustrate the combined effects of U_g and screen/film unsharpness. They indicate that it is important to evaluate the effect of U_g in chest radiography so that the screen/film system is properly selected and used in terms of total system sharpness. Obviously, the smallest focal spot with the sharpest screen system may not be practically possible for many chest radiography techniques due to limited tube output and excessively long exposure times which may be required. However, we are of the belief that the effects of U_g and screen/film system unsharpness as presented for various planes in the chest offer important, but often overlooked, information which can be used to optimize chest radiography imaging techniques.

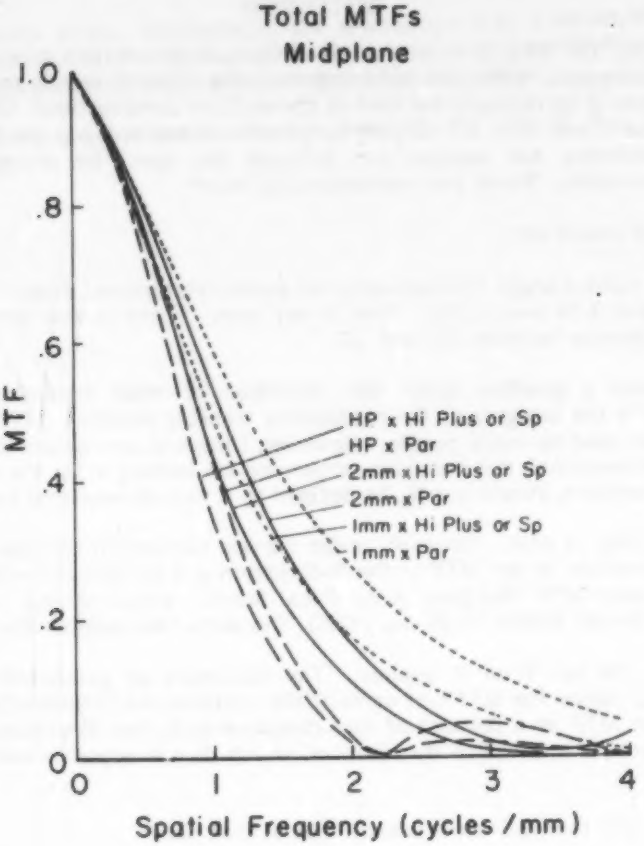

Figure 4

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DISCUSSION

DR. NICKOLOFF: Dr. Rao, when you spoke of the quantum mottle decreasing as the speed of the screen increased, were you assuming that the type of screen remained the same? What would happen if we changed the kind of screen/film combination? Obviously, what you are saying is true if you have all calcium tungstate screens and you are changing the speed by changing thickness, but suppose you increase the speed by changing from calcium tungstate to rare-earth. Would the expression still hold?

DR. RAO: No, it would not.

Also, let me make a slight correction to my earlier statement. I said "The constant that you saw there was 2.30 over 6.93." That is not true. There is also another factor, 1.75, which is the difference between 2.0 and .25.

DR. DOI: I have a question about the definition of noise equivalent passband (N_e), which is related to the integral of the modulation transfer function (MTF). Of course, this constant has been used by many people. However, I think as you define it in your slides the integral is one-dimensional, but the noise pattern we are looking at on the radiograph is two-dimensional. Therefore, should not N_e be defined by a two-dimensional integral.

DR. RAO: Yes, that is true. However, under certain circumstances, you can multiply the MTF in the X-direction by the MTF in the Y-direction and the product will be equivalent to the two-dimensional MTF (Burgess, A.E.: Focal Spots. I:Separability, II:Models, III:Field characteristics. Invest Radiol 12:36-61, 1977.) The same idea applies also to N_e.

DR. DOI: No, I do not think it applies. The discussion of geometric unsharpness is a separate subject. Since the MTF's of screen/film systems are rotationally symmetric, you cannot obtain the MTF as a product of two components in two directions--except in some special cases, such as a Gaussian distribution, which do not apply to common screen/film systems.

DR. RAO: What will the equivalent passband be in that case?

DR. DOI: I do not know, but the result will not be the same.

DR. METZ: I was quite independently going to ask the same question as Dr. Doi. If you have a rotationally symmetric MTF, then the appropriate two-dimensional integral becomes simply 2π times a single integral of frequency times MTF^2. Now, if the MTF shape is constant and just changes in width, then both that integral and that result are always proportional to the square of the MTF width, and inversely proportional to the square of the line spread function (LSF) width or point spread function (PSF) width. But the trick of considering PSF to be the product of two one-dimensional LSF's works for rotationally symmetric MTF's only if the MTF is Gaussian; otherwise, your "trick" does not work, and it can yield a very wrong quantitative result. I agree with Dr. Doi that if you are going to be comparing two systems which have PSF's or MTF's of possibly different shapes, then in principle you must use a two-dimensional integral of MTF^2 or, for rotationally symmetric point spread functions, a one-dimensional integral of frequency times MTF^2.

DR. RAO: Yes, I am aware of the complexities you are bringing up. However, let me pose another question to you. If the screen/film system MTF can be approximated by an exponential function, then is it valid to say that the two-dimensional N_e is given by the expression I quoted?

DR. METZ: If one assumes that all MTF's of interest are exponential, then your approach works in the sense that the ratio of two of your results will equal the ratio of the two actual two-dimensional N_es. However, your approach does not give the correct quantitative N_e, and the error depends on the shape of the MTF; so, in general, your approach cannot be used to determine N_e in absolute terms and it cannot be used to determine relative N_e values for systems with

MTF's of different shape. Incidentally, the assumption that a screen/film system MTF is exponential--that is, of the form exp (-af)--is only a rather crude approximation, and different systems can have MTF's of somewhat different shape.

DR. RAO: That assumption is inherent even in the use of F_{50} and F_{10} as single measures.

RISK/BENEFIT CONSIDERATIONS IN CHEST RADIOGRAPHY

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INTRODUCTION

It is this author's opinion that while the risks associated with radiographic procedures are reasonably calculable, the benefits are very personal and depend mostly on individual appraisal; hence, they will not be discussed here. However, before considering the risks associated with chest radiography per se, it is probably best to review the biological effects of radiation (Hall, 1978; Fabrikant, 1972) in order to place the various risks in proper perspective.

Very simplistically, the absorption of ionizing radiation by human tissues can result in four distinct effects: cell sterilization, production of choromosome aberrations, production of mutations, and finally, production of cancer. The magnitude and importance of each effect as well as the time interval between exposure and appearance of the effect will of course depend on the particular tissues being irradiated. For example, it is possible to have a cell receive just enough radiation to stop division yet not affect cellular functions such as the production of key enzymes. No effect of the radiation will then be observed until the cell attempts division at which it must fail. In a few cases, the nucleus may divide inside the cell membrane producing a multi-nucleated cell which ultimately will die and lyse away. Also, a cell may actually complete the first division but fail to divide during the second generation of divisions. This happens in a very small (negligible) fraction of the cases. Normal (unirradiated) cellular interdivision times may be as short as 10 hours in the case of rapidly dividing cells or months and years in the case of nerve cells. In some cases, the sterilizing effect of radiation may be delayed until the tissue is challenged to reproduce as in the case of damage to the skin. Radiation can also create a nidus for rapidly dividing cancer cells which may not be detected for up to 20 years due to the large number of cells required for detection or a limited growth rate.

CELL STERILIZATION

Cell death or sterilization is the most likely effect of radiation. Measurements have shown that cells in the rapidly proliferating tissues in the body (e.g., the reticuloendothelial cells with an LD_{50} of 150 rads) have a probability of about 0.004 of being sterilized by 1 rad of radiation. This means that if a million cells receive 1 rad of radiation, on the average about 4,000 will be killed. However, if the cells have a doubling time of 10 hours, the remaining 996,000 cells will produce the missing cells in about 3 minutes. Considering this and the enormous number of cells dying from natural causes every minute in the body, it may be seen that the effect of cell sterilization from 1 rad is quite negligible. One possible exception to this is the production of cataracts in which the dead cells are retained and accumulated in the eyes over many years. Many hundreds of roentgens are usually required with an average latent period of 8 years to produce a visible opacity in the lens. Whole organs can also be affected by cell sterilization but many ultimately recover through replacement. It has been demonstrated that radiation doses of 400 rads to 500 rads to human gonads will not necessarily induce permanent sterility but rather temporary infertility followed by a return of fertility. Contrariwise, the tissues in the developing embryo and fetus are quite sensitive in that exposures on the order of 10 rads may produce a child with some form of abnormal organ development in perhaps 5 percent of the live births. Obviously, there is a limited supply of cells available, each with key information, and the killing of a few critical cells at a particular time or the production of transformed cells affects the further development of the child. The organs that are affected are those under development at the time of exposure (Rugh, 1964).

CHROMOSOME ABERRATIONS

When cells are irradiated, "breaks" may subsequently be observed in the chromosomes. Although no direct evidence is available, many investigators feel that such gross chromosome aberrations are directly related to and possibly responsible for lethality. Further, these aberrations are not unique. Chromosome "breaks" are not only observed in normal tissue cells but no aberration has been produced by radiation that has not also been seen in unirradiated controls. Measurements have established that the probability of producing a break is .005 per rad which is very close to the probability of producing cell death (Elkind and Whitmore, 1967). Furthermore, variation of those parameters that change the sensitivity of cells to the lethal effects of radiation (e.g., oxygen consumption) also affect in a similar manner the probability of producing aberrations. It is highly probable that the large visible chromosome defects discussed here do lead to sterility; however, smaller and unobservable effects may also occur that do not lead to cell death but which may change the inheritable characteristics of the cell. This resulting cell could then constitute a mutation of the original cell line.

PRODUCTION OF MUTATIONS

For missense mutants of mammalian cells, the genetic damage from a mutagenic agent is heritable and should be revertible by further treatment with a secondary mutagen. Those cells for which reversion has not yet been proved are usually classed as variants but could be deletions (which do not revert) or double mutations. Interestingly, no mutagenic agents have been found to produce a mutation that has not been observed in the normal population. They merely increase the rate of production of such changes. Because of the difficulty of identifying mutational changes in mammalian cells per se, mutations observed to date have been limited to those affecting the basic biochemical processes in the cells. Examples are induced resistance to radiation which inolves the utilization of oxygen and repair mechanisms, increased resistance to IUdR, induced resistance to 8-azaguanine, and production of glycine deficiency. The probability of producing such a mutation of a mammalian cell averages about one in a million per rad which is about 1/4000 of the probability of inducing sterilization.

If now the gonads of mammals--including man--are irradiated, any mutated germ cell or ovum can unite with its mate during conception and ultimately produce an aberrant offspring. Either or both type cells can have been irradiated with the probability of producing a mutation doubled if both are exposed. Russel and others (Russel, 1968; Lyon et al., 1972) have scored the mutations appearing in the progeny of irradiated mice and found a probability of two in ten million per rad of producing any one specific mutation when irradiating either partner. Both male and female mice seemed to be equally sensitive to the mutational effects of radiation. An additional observation was that if more than 7 weeks intervened between irradiation of the female mice and conception, the number of mutations dropped to zero--implying of course complete repair of the genetic damage. This was not seen in the males. This effect has not yet been observed in humans but most likely occurs. Rather limited data on radiation induced mutations in humans (National Academy of Sciences, 1972) indicates that the amount of radiation required to double the natural mutation rate is about 80 rads to either partner. Since the natural mutation rate is 9 aberrant offspring per 100 individuals (Trimble and Doughty, 1974), we then calculate a probability that 1 rad to either mate will produce about one mutation in a thousand conceptions. This probability will be much reduced if only the female is irradiated and more than 7 weeks have elapsed between irradiation and conception. Since there are about one thousand suspected dominant mutations in humans that can occur, it is obvious that the probability of producing any one particular mutation (single locus) in humans is quite similar to that observed in mice.

CARCINOGENESIS

The fact that radiation can induce cancer has been known since 1896; however, the basic cause is still unknown. Fortunately (or unfortunately), data exist from past tragedies that allow an estimate of the probability of radiation carcinogenesis (National Academy of Sciences, 1972). These are shown in Table 1. The first point to note is that these data are valid only if the whole organ is irradiated and the dose is averaged over the whole volume of the organ. For example, if only one breast were exposed to a uniform dose, the probability (per rem) of carcinogenesis would be only one-half of that listed. Secondly, the listed probability of induction is only for fatal cancers. For the probability of simply inducing a tumor in a given organ without regard to fatality, the listed probability should be divided by one minus the fractional 5-year survival. However, in this discussion, we shall be concerned only with the induction of fatal malignancies with the understanding that the mere induction of tumors will be about twice as large since the cure rate averages about 50 percent for all tumors. Lastly, it is to be pointed out that for any one exposure, the latent period can be as long as 25 years. Thus, if an individual receives an acute exposure of 1 rem to the bone marrow, the probability of the occurrence of a fatal leukemia is about one per million per year for the following 25 years totaling 25 per million in a lifetime. This conclusion is consistent with National Academy of Sciences (1972) data which indicate that 1 rad of whole body irradiation will produce between 60 and 180 fatalities from malignancies over the lifetime of a million individuals. Regarding dose-rate effects, the above data on radiation carcinogenesis were obtained with the dose rates ranging from the very intense blast of an A bomb through normal therapeutic and diagnostic procedures to low dose rates from internally deposited radionuclides. With few exceptions, all dose-effect relationships were best fit by a straight line that extrapolated through zero. Further, the doses ranged from a few to many thousands of rem. We shall thus assume linearity of effect with absorbed dose and the absence of dose-rate effects. Interestingly, recent analysis of all available data on irradiation of children in utero showed that the risk of inducing a malignancy was approximately the same as that for adults (Pochin, 1976). While many scientists feel that carcinogenesis is most likely due to radiation induced mutations, it is to be pointed out that the probability of producing cancer is on the order of 10^{-15} per rad per cell when one accounts for all the cells in each organ. Since the probability of mutagenesis in proliferating mammalian cells is on the order of one per million per rad per cell, the relation betwen the two effects is far from clear. For multiple exposures, to a first approximation one may simply add the individual probabilities involved for each procedure. However, for greater accuracy, one must account for the fact that the average lifetime of an individual is longer than the 25 year latent period and that the exposures are usually spread over a long time (Gregg, 1977). Simple addition obviously produces a "worst case" calculation.

CHEST RADIOGRAPHY

Surveys (HEW, 1973) show that most chest radiographs are taken at 72 in with a 14 in x 17 in field size and a tube potential between 55 and 95 kVp. The average skin dose was 44 millirem per exposure and 53 percent of all examinations were single exposures (mostly PA) while 41 percent used two films (PA and LAT) and 3 percent used three films. On the average in the U.S., approximately 30 out of every 100 individuals had a radiographic chest examination producing about 65 million examinations in one year. Considering the variation in the number of films per examination, we shall assume for lung tissue calculations that the average skin exposure is 63 millirem at 75 kVp and 72 in. Following previously described absorbed dose calculations (Gregg, 1977), we shall also assume an absorption coefficient $\mu = 0.07\ \text{cm}^{-1}$ for lung tissue under these conditions with the average lung having an effective solid tissue thickness of 5 cm. This produces a ratio of average absorbed to incident skin dose of $\alpha = D_{av}/D_o = 0.86$ which in turn produces an average absorbed lung tissue dose of 54 millirem. For the bone marrow, calculations and measurements (Rosenstein, 1976) show an absorbed dose of 60 millirem (PA), 30 millirem (AP), and 24 millirem (LAT) per 1,000 millirem skin dose. This leads to an average bone marrow absorbed dose of 4.3 millirem. It must also be remembered that the marrow exposed during a chest radiograph constitutes only about 25 percent of all the body marrow. Exposure to the thyroid during a chest radiograph averages about 235 millirem (AP), 18 millirem (PA), and 81 millirem (LAT) per 1,000 millirem skin dose. This produces an average thyroid dose of 2.1 millirem per examination. For the male, the testicular dose is negligible while the female will receive an average ovary dose of about 0.3 millirem per 1,000 millirem skin dose or 0.02 millirem per examination. One last organ to consider is the female breast which is quite sensitive to radiation carcinogenesis. As shown recently (Nash et al., 1979), the chest examinations described above will lead to an average absorbed breast dose of 4 millirem per examination for PA exposures. If AP should be used, the breast (and thyroid) will be increased considerably. Calculations for bone, stomach and upper GI tract are not presented because the probability of carcinogenesis is quite small relative to the organs discussed above.

Table 1. Cancer statistics*

Primary site	Prob. of occurrence ($\times 10^4$/yr)	Fractional 5-yr survival (F_a)	Prob. of fatal malignancy ($\times 10^4$/yr)	Prob. of induction (Pra) ($\times 10^6$/yr/rem)
Marrow (Leukemia)	1.0	0.13	0.87	1.0
Bone	0.1	0.20	0.08	0.2
Brain (& CNS)	0.5	0.30	0.35	1.0
Lung	3.8	0.08	3.50	1.0
Female breast				
Ages 20-34	1.1	0.60	0.44	6.0
Ages 35-49	10.4	0.60	4.20	2.0
Thyroid	0.4	0.80	0.08	1.0+ 6.0**
Colon	3.0	0.34	1.98	
Stomach	1.0	0.09	0.91	
Tongue	0.2	0.27	0.15	1.0†
Liver	0.2	0.30	0.14	
Pancreas	0.9	0.02	0.88	
Prostate	2.0	0.40	1.20	

*Averaged for all races, sexes, and ages except as noted
+Adults
**Children
†For total of remaining tissues

Table 2 lists the increase in various risks to an individual having only one chest radiograph. It is to be noted that with the exception of thyroid cancer in children, 100 such radiographs of the same individual would still produce risks of carcinogenesis less than 1 percent of the spontaneous rate. Thus, it would seem that most chest radiographic examinations are quite innocuous.

The preceding discussion has been concerned only with risks to the individual. Regarding the benefits, these should be evaluated by the knowledgeable patient and his advocate. To some individuals, the mere alleviation of continual pain is worth almost any risk of malignancy, to others the opposite may be true. Participation in surveys should also receive serious consideration. For example, at what level is the risk of indicing a malignancy worth the early detection of same? This carcinogenic risk/benefit ratio can be calculated for such surveys but few individuals wish to be associated with a recommendation that implies the production of a small but finite number of malignancies in a given population. Since such cancers are produced naturally, it is a human tendency to blame the diagnostic procedure for any one or all of them.

Table 2. Risks due to one chest radiograph

Effect	Spontaneous rate (per yr)	Induced rate (per yr)	Increase in risk (%)
Leukemia	10^{-4}	10^{-9}	.001
Lung cancer	2×10^{-4}	5×10^{-8}	.025
Thyroid cancer			
Adults	0.4×10^{-4}	2×10^{-9}	.005
Children		1.2×10^{-8}	.030
Female breast cancer			
Ages 20-34	1×10^{-4}	2×10^{-9}	.002
Ages 35-70	1×10^{-3}	8×10^{-9}	.0008
Genetic defects in offspring	.09	2×10^{-8}	2×10^{-7}

In 1968, about 40 million chest fluorographies were carried out in Japan. The results of such a survey are presented in Table 3 (Kitabatake, 1973). Not shown are calculations indicating the possibility of inducing 46 leukemias by the radiation exposures which should be compared with 97,000 spontaneous leukemias. Also estimated were 7 possibly radiation induced lung cancers and 150 estimated genetic deaths over the first 10 generations. Since the incidence and mortality rates for tuberculosis have diminished considerably in Japan since 1940, it is widely accepted in Japan that mass chest fluorography has made a great contribution to the prevention and control of tuberculosis. Of interest is the fact that the average marrow dose in these chest exposures was estimated to be about 35 millirem which is to be compared with 4 millirem in the United States as listed previously.

Table 3. Results of chest radiography of 40 million people in Japan in 1968

Disease detected	Number detected
Curable TB	38,629
Curable lung cancer	758
Incurable TB	5,818
Incurable lung cancer	2,271
Benign pulmonary disease	1,000,000

There are also other societal benefits to be considered in terms of monies accruing from workdays gained by avoidance of illness relative to the actual cost of the 65 million chest examinations performed annually in the U.S. Preliminary calculations would indicate enormous benefits to society from currently prescribed radiographic chest examinations without consideration of possible malignancies induced in the population (National Academy of Sciences, 1977).

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EXPOSURE IN CHEST RADIOGRAPHY - "NEXT" RESULTS

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INTRODUCTION

This session of the Chest Conference addresses patient exposure in chest radiography. The Nationwide Evaluation of X-Ray Trends (NEXT) system can contribute a significant amount of information related to chest exposures. In addition, the Bureau of Radiological Health has other data sources which will be discussed.

These additional data sources include the X-Ray Exposure Studies of 1964 and 1970 (XES 1964 and XES 1970), which were comprehensive surveys of United States population exposure to x rays (1, 2). The Bureau is presently developing a survey system that will provide updated information on examination rates; however, data from XES 1964 and XES 1970 remain useful.

The NEXT program is a cooperative effort of the Bureau and State radiological health programs to estimate exposures to a standardized patient for several common radiographic procedures. The NEXT survey for chest specifies that the x-ray machine be set up for a single P/A chest projection on a patient with a 23-cm thorax. Details of this system have been described previously (3). Current NEXT survey procedures do not include photofluorographic or automatic exposure control units.

NEXT data coleection began in 1972 with a few States participating. Currently, 42 States submit data for inclusion in the NEXT system. The first 5 years of NEXT "All States" data, from 1973 through 1977, include surveys of 6,631 P/A chest projections.

The basic NEXT survey does not include information on image receptor systems used. An Image Receptor Module (IRM) was developed as an optional addition to the NEXT system in 1976. This module provides additional information on type of film and screen, use of grid, and details of processing. To date we have this information for 620 NEXT surveys of chest projections.

The NEXT exposure data presented in this paper are weighted to represent population exposure. This is accomplished by taking into account the workloads of the machines surveyed as well as the proportion of facilities surveyed within each State. The smaller amount of IRM survey data have not been weighted. The data are indicated in tables and histograms as "percent of workload" for weighted data and "percent of surveys" for unweighted data.

POPULATION EXPOSED

Examination frequency studies show that chest x rays are the most common medical x-ray examination, accounting for 46 percent of all medical radiographic exams performed in 1970. As Table 1 shows, 76 million persons had medical radiographs, of whom 48 million had chest radiographs. This latter group had 49 million radiographic chest exams, plus 10 million photofluorographic examinations.

Table 1. Estimated annual medical x rays by numbers of persons exposed and numbers of chest examinations

PERSONS EXPOSED*	NUMBER IN MILLIONS
MEDICAL RADIOGRAPIC	76
CHEST RADIOGRAPHIC	48
CHEST EXAMINATIONS	**NUMBER IN MILLIONS**
RADIOGRAPHIC	49
PHOTOFLUOROGRAPHIC	10

*INCLUDES 9.7 MILLION PERSONS HAVING PHOTOFLUOROGRAPHIC EXAMINATIONS.

(XES 1970)

Table 2 shows a distribution of chest x-ray examinations by age. The breakdown of annual exam rates per 100 persons shows a general increase with age. Among individuals 45 and over the rate is nearly 50 exams per 100 persons.

Table 2. Estimated annual distribution and rate of chest x-ray examinations by age (includes photofluorographic)

AGE	PERCENT OF CHEST EXAMS	RATE PER 100 PERSONS
UNDER 15	9	9
15-29	22	27
30-44	22	39
45-64	32	46
65 AND OVER	15	46

(XES 1970)

Various extrapolations of these data have been made to estimate current numbers of examinations. If one assumes that the chest exam rates remained constant from 1970 to 1979, and adjusts for the changing population age distribution, there would be approximately 68 milion chest exams this year. A linear extrapolation of numbers of chest exams from XES 1964 and 1970 would estimate 74 million radiographic chest exams in 1979. An exponential extrapolation would project 79 million chest exams.

EXPOSURES AND DOSES

The range of exposures measured by NEXT over the period 1973 to 1977 is presented in Figure 1. This histogram shows exposure at skin entrance (ESE) in mR, free in air. The exposure ranges from less than 5 to more than 100 mR for a single P/A chest projection. The variation of technique factors, image receptor systems, and grid use contributes to this

wide range of ESE. The average value for the NEXT chest data is 21 mR, while the median is 16 mR. The difference reflects the shape of the curve, the tail of high exposures having more effect on the mean than on the median.

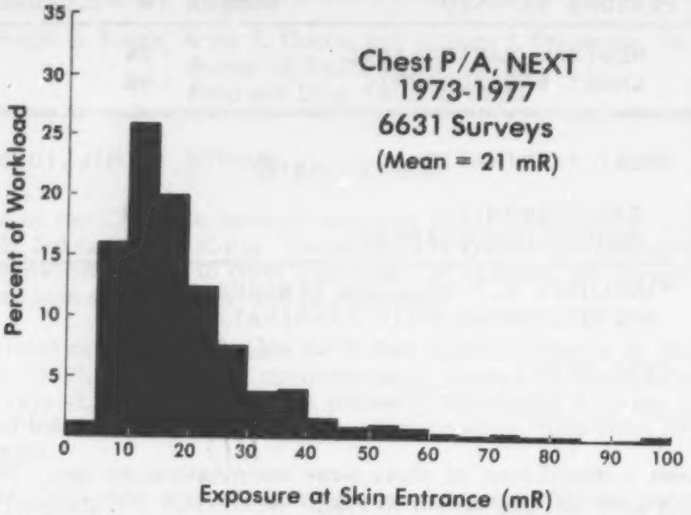

Figure 1. Frequency distribution of chest exposures.

The average NEXT values have not changed much since 1973, as shown in Figure 2. XES 1970 showed an average ESE for the P/A chest of 25 mR. As a rule of thumb, a lateral chest will require three times the exposure of the P/A.

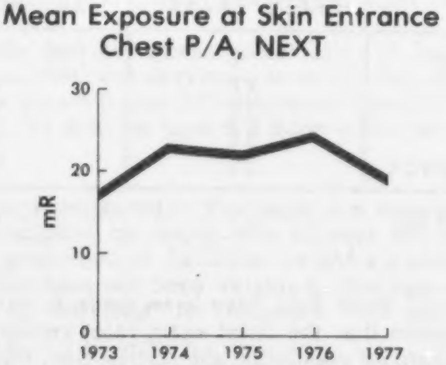

Figure 2. Mean exposure at skin
entrance by survey year.

Table 3 shows the calculated mean values of organ doses produced by P/A chest examinations as measured by NEXT. The values are quite small, ranging from less than 0.5 mrads for gonad doses to 1.9 mrads for red bone marrow dose. Although these individual doses are small, the total dose to the population must be considered because of the large number of these exams performed each year.

Table 3. Mean organ doses for radiographic P/A chest projections.

BODY ORGAN	CALCULATED DOSE (MRAD)
OVARIES	< 0.5
TESTICLES	< 0.5
RED BONE MARROW	1.9
THYROID	0.5

(NEXT 1973-1977)

FACILITIES AND EQIPMENT

The XES 1970 data presented in Table 4 show the type of facility in which chest examinations were performed. At that time, two-thirds of radiographic chest exams took place in hospitals. It is interesting to note that while the "Health Agency and Other" category accounted for only 9 percent of radiographic chest examinations, 90 percent of the photofluorographic chest examinations were performed in this category of facilities.

Table 4. Distribution of radiographic chest examinations by type of facility.

TYPE OF FACILITY	PERCENT OF EXAMS
HOSPITALS	67
PRIVATE OFFICES AND GROUPS	24
HEALTH AGENCIES AND OTHERS	9

(XES 1970)

Regarding x-ray equipment used for chest radiographs, NEXT found that 14 percent of x-ray units surveyed used circular collimation, with the remainder rectangular; however, units with circular collimation accounted for less than 10 percent of the chest workload (Table 5). Twelve percent of the machines surveyed were portable, but they accounted for only 3 percent of the P/A chest workload in NEXT.

Table 6 shows that chest films were processed manually in about 28 percent of the 620 Image Receptor Module surveys. Three percent of the hospitals and 62 percent of the private offices included in the IRM survey used manual processing.

TECHNIQUE FACTORS

The source-to-film distance, x-ray tube potential, half-value layer, and use of grid are technique factors of interest in analyzing chest radiography. NEXT indicates that about 95 percent of P/A chest projections were done at a source-to-film distance (SFD) of 72 inches. Only nine out of 6,631 surveys recorded a 120-inch SFD. One third of the surveys on portable units showed an SFD of approximately 40 inches. However, a larger proportion, about 40 percent of the surveys on portable machines, recorded an SFD of 72 inches.

Table 5. Distribution of P/A chest workload by type of equipment.

TYPE OF BEAM-LIMITING DEVICE	PERCENT OF WORKLOAD
CIRCULAR	9
RECTANGULAR	91
TYPE OF X-RAY MACHINE	**PERCENT OF WORKLOAD**
STATIONARY	97
PORTABLE	3

(NEXT 1973-1977)

Table 6. Distribution of P/A chest film processing by type of facility.

TYPE OF FILM PROCESSING	ALL FACILITIES	HOSPITALS	PRIVATE PRACTICE	OTHERS
AUTOMATIC	72%	97%	38%	83%
MANUAL	28%	3%	62%	17%

(IRM 1976-1979)

NEXT data show that 99 percent of P/A chest projections were done with 14 in x 17 in film (Table 7) and about 1 percent with 14 in x 14 in film.

Table 7. Distribution of P/A chest workload by film size.

FILM SIZE (INCHES)	PERCENT OF WORKLOAD
14 X 17	99
14 X 14	1

(NEXT 1973-1977)

Table 8 shows that in 28 percent of the IRM surveys a grid was used for an average-size patient (23-cm thorax).

Figure 3 shows the distribution of NEXT data by peak tube potential (kVp). Most of the projections (62 percent) were performed with a kVp between 70 and 85; a few (14 percent) appear in the 100, 110, and 120 kVp categories. The kVp data from the IRM surveys can be broken out by grid use (Fig. 4). (Note that in this figure each of the histograms is plotted as

a percentage of the number of surveys in its respective category, and that the actual number of surveys with grid is considerably less than the number without grid.) As expected, more of the techniques above 100 kVp included use of a grid, although many of those using lower kVp also used a grid.

Table 8. Distribution of P/A chest projections by grid use:

GRID USED	PERCENT OF SURVEYS
YES	28
NO	72

(IRM 1976-1979)

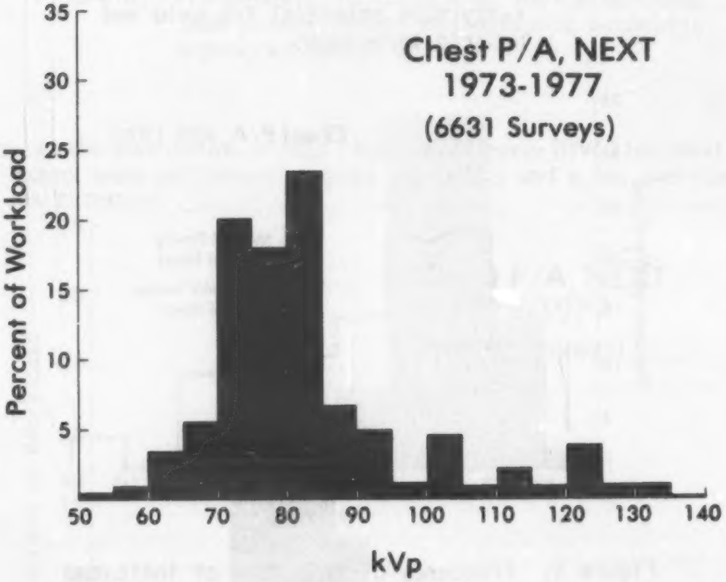

Figure 3. Frequency distribution of indicated x-ray tube potential.

It is also interesting to examine kVp by the type of exposure timing used. NEXT does not have procedures to survey automatic exposure control (AEC) units, but the XES 1970 study collected data on these machines. Figure 5 shows the kVp associated with automatic and manual machine timing. Note that automatic exposure control units accounted for 10 percent of the total chest films in 1970. It can be seen that facilities with AEC units used higher kVp than those with manual units.

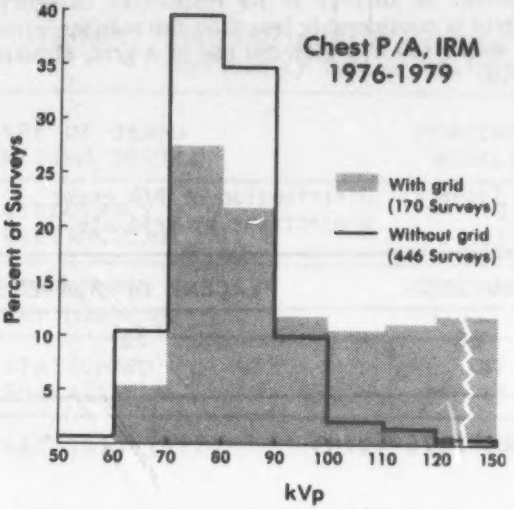

Figure 4. Frequency distribution of indicated x-ray tube potential for grid and nongrid techniques.

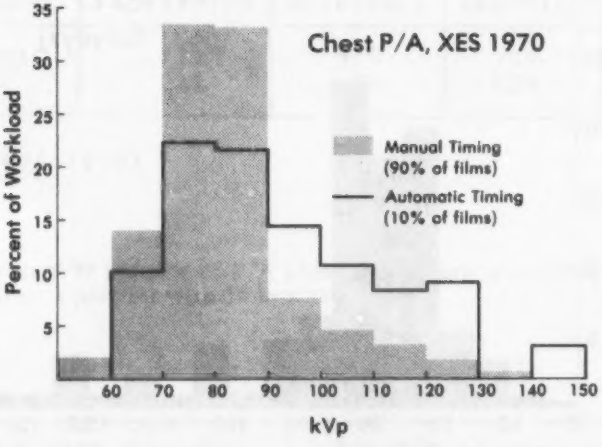

Figure 5. Frequency distribution of indicated x-ray tube potential for manual and automatic exposure control units.

The Bureau is currently performing a special survey of automatic exposure control units. Figure 6 shows data from 72 surveys of units used for the P/A chest projection. Note the predominance of kVps. Incidentally, grids were used in 85 percent of these cases, again indicating a high correlation of grid use with high kVp. While these data may be biased by inclusion of many new x-ray machines, they again show that NEXT chest data taken only from manual machines may not accurately represent the total practice of chest radiography in the United States. Data are needd both on relative use of automatic exposure control units for chest examinations and on patient exposures resulting fom the use of AEC.

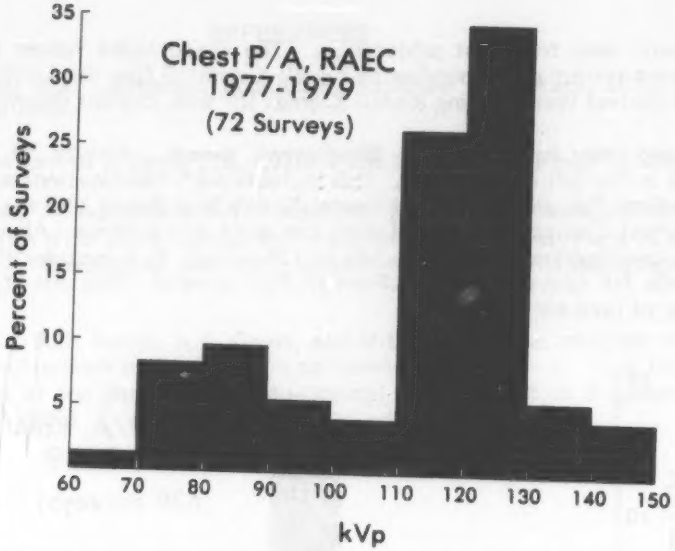

Figure 6. Frequency distribution of indicated x-ray tube potential for radiographic automatic exposure control units.

Figure 7 shows the distribution of NEXT half-value layers (HVL) for chest examinations. Some of the exams were performed at quite high HVLs and a few were performed with unacceptably low filtration.

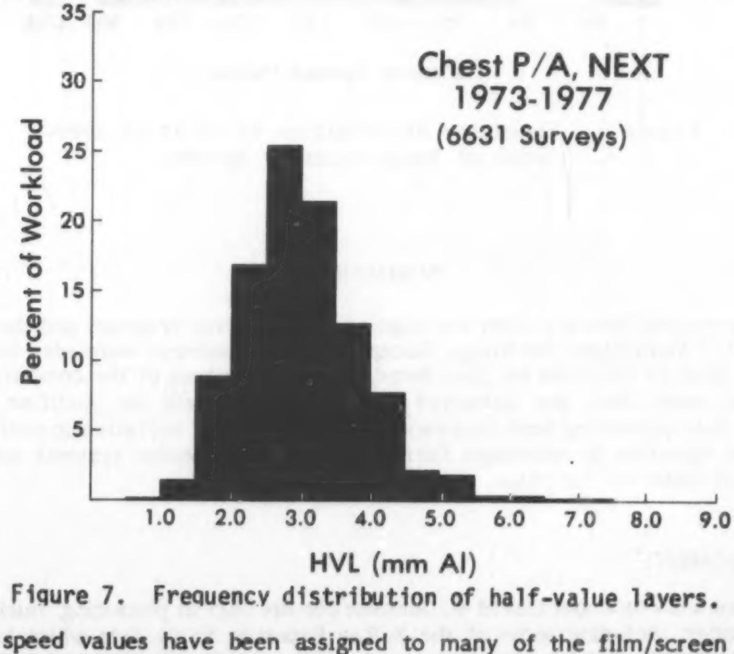

Figure 7. Frequency distribution of half-value layers.

Relative speed values have been assigned to many of the film/screen combinations reported in the IRM surveys. This assignment has been described previously (4). Indexes are relative to the combination of Kodak RP film and DuPont Par-Speed screens, which was assigned a speed index of 100. In the 620 IRM surveys, 125 different film/screen combinations were used for chest projections. The speed index values ranged from 53 to 305, the slowest system corresponding to Kodak X-omat G film used with Kodak Regular screens, and the fastest system being Kodak X-omat RP with DuPont Quanta II screens.

Figure 8 shows the distribution of film/screen speeds. Nearly 50 percent of the combinations fall in the 100 to 125 range. This includes such combinations as Kodak X-omat RP film with DuPont Par and Hi-Speed screens, Kodak Blue Brand film used with the same screens, and DuPont Cronex 4 film used with the same two screens. About 20 percent of the film/screen combinations fall in the 150 to 175 range. This includes DuPont Cronex 4 and Kodak X-omat RP film used with DuPont Hi-Plus screens. Only one of the 620 surveys indicated the use of rare-earth screens.

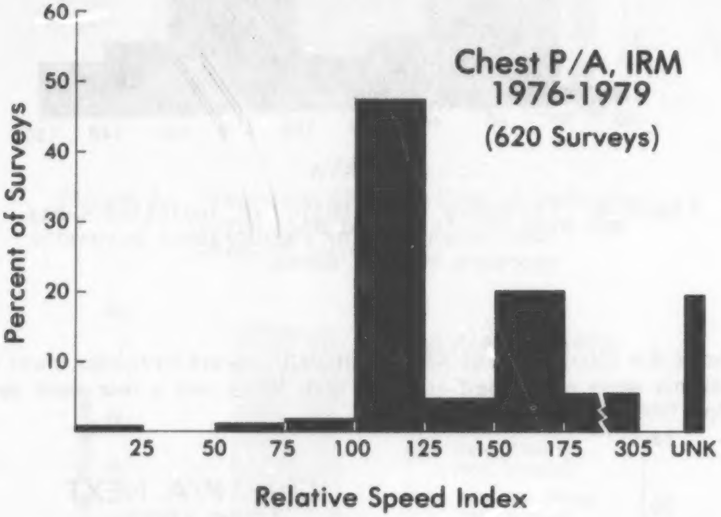

Figure 8. Frequency distribution of relative speed index of image receptor system.

SUMMARY

The data presented here are from the ongoing NEXT survey program and the 1970 X-Ray Exposure Study. Data from the Image Receptor Module surveys were also included. The results of the IRM survey must be considered tentative because of the comparatively small data base; as more data are collected the conclusions will be modified or verified. However, the data presented here illustrate the magnitude of variation in patient exposure as well as the variation in technique factors and image receptor systems used for chest examinations at different facilities.

ACKNOWLEDGMENT

The authors wish to thank David W. Johnson for his help in producing much of the data used in this paper, including some of the X-Ray Exposure Study data which have not been previously published in this format.

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EXPOSURE SURVEY IN CHEST RADIOGRAPHY

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The skin exposures from postero-anterior and lateral chest radiographs have been measured in several hospitals and private offices in Toronto. From one x-ray room to the next, exposures differed by a factor of up to 10. The reasons for these differences are examined and lie in the selection of screens and films, in the kVp employed, in the use of a grid or an air gap technique, and in the development of the film. These factors should therefore receive priority in any attempt to reduce the higher doses toward the lowest values.

From the factors in the techniques studied, an optimum chest technique is proposed which should result in a skin exposure of 5 to 20 mR for a PA radiograph on all patients of average size and of 25 to 40 mR for a lateral radiograph. At these exposures the risks, although minimal, are probably greatest for leukemia in the male and for leukemia or cancer of the right breast in the female.

METHODS

The techniques, procedures and performance of the equipment were studied in some detail in each x-ray room. Kilovoltage, milliamperes, time, x-ray wave form and system geometry were noted. The type of grid, type and age of screens and the type of film were also noted and film processing was monitored by sensitometry.

Skin exposures were measured with a calibrated dosimeter either on patients or on a phantom which simulates an average male chest, described by Hughes and Schulenburg (1). In some hospitals repeat rates were measured and the reasons for high repeat rates studied.

Difficulties in carrying out chest radiography were discussed with the x-ray technologists, and image quality was discussed with radiologists.

RESULTS

RANGE OF EXPOSURES

For the average adult, exposures for the PA view were from 8 mR to 80 mR, with the peak of the distribution around 20 mR. For the lateral view the exposures were fom 20 mR to 200 mR, with the peak of the distribution around 50 mR. The ratio of lateral to PA exposure was about 2.5 in most cases.

This distribution of exposures is similar to that found in the NEXT survey, but lacks some of the extremely high values found in that survey (2). It is also similar to the distribution of exposures found in a survey in Manitoba (range 5 to 85 mR) although the peak of that distribution is at 15 mR, rather lower than in Toronto (3). However, the measurements in Manitoba were carried out with a phantom (9 cm water equivalent) which may simulate a rather thin adult. In both Ontario and Manitoba high exposure photofluorography is virtually eliminated.

REASONS FOR EXPOSURE DIFFERENCES

Table 1 shows the reasons for exposure differences in our small sample. All the machines that we investigated contained adequate filtration and this was not a significant factor in the exposure differences. There were four main factors and Table 1 gives the ratio by which exposures differed because of these factors. These factors interact, for instance, when a high kVp technique with air gap and no grid is used, and the ratios cannot simply be multiplied together to find the total exposure reduction that is possible. However, the table does indicate where priority should be placed in attempts to reduce exposure. Because it does not appear in Table 1, filtration should be added to the list of items to check.

Table 1. Reasons for exposure difference

screen film speed	4:1
kVp (80 to 140)	2.5:1
with or without grid	2:1
film development	2:1

UNNECESSARY EXPOSURE

In addition to the differences in exposures noted above, there was a significant amount of unnecessary exposure. Three examples will be considered. They are typical of the practical problems that may increase dose to the patient.

a. In one room, two PA views were taken on each patient, displaced to the left and to the right. This was because the automatic film changer was causing severe static marks about 3 cm wide down each edge of the film and these marks were interfering with the edges of the lung fields. This problem had lasted some weeks at the time of our study and a solution was still being actively sought.

b. Previous studies showed that repeat rates for chest radiography varied between 5 percent and 15 percent. These figures were found for both manual timing and phototiming, and 3-5 percent was due to errors of positioning and movement; the rest were due to incorrect density. With manual timing the repeat rate depends upon the experience and care of the technologist, but also on the consistency of exposure of the x-ray machine. With phototiming on some machines, several factors, which will be considered below, led to high repeat rates and in some cases to the abandoning of phototiming in favor of manual control. However, at least three properly functioning and well set up phototiming systems produced excellent results, with repeat rates at or below 5 percent.

c. In several instances, changes in film speed of 20 to 40 percent have been found to occur due to permanent changes in manufacturing and the film characteristic. Such changes lead temporarily to high repeat rates and these may be blamed on the x-ray equipment rather than on the film. In a typical case a doubling of the repeat rate occurred for about 2 months.

While these unnecessary exposures should be elminated if possible, the unnecessary cost that they entail is of greater impact than the increased patient exposure. Reduction of these unnecessary exposures could lead to a 10 to 30 percent reduction in patient dose while selection of a faster imaging technique might reduce dose by a factor of three.

FIELD SIZE

The size of the x-ray field relative to the size of the film was not studied in detail; however, it was observed that the collimation, when done manually, was usually to within 1 inch of the film size on all sides.

Collimation was automatic on several machines, and was close to the film size in all. However, in one machine poor design of the collimating aperture led to a large penumbra about 8 cm wide all around the film area. This increased the integral dose to the patient by about 30 percent, but was easily removed by a re-design of the collimator.

PHOTOTIMING

The failure of phototiming to operate satisfactorily is largely due to the short exposure times which are required for chest radiography. Many x-ray generators have been found to give inconsistent exposures for high tube currents and short times, even though they are very consistent for longer times. Some of these problems can be solved by using lower mA and longer times, on the order of 20 ms. Indeed, manual techniques often use exposure times of 25 to 50 ms. However, this may lead to greater motion blurring.

If it is desired to use very short times, then forced extinction of the exposure can give exposure times as low as 2 ms. However, the stored energy in the high tension circuit may lead to a substantial x-ray exposure after the exposure is terminated by the phototiming circuit. Compensation for patient thickness may therefore be poor. This problem is greatest for low tube mA, but may still be significant at high tube mA.

In one chest unit the lateral phototiming sensors were found to be too high and were partly above the apexes of the lungs. Incorrect positioning of the phototiming sensoring areas is a common cause of trouble with phototiming in all areas of the body.

Phototiming, when properly set up and adjusted, gives very consistent radiography with very low repeat rates. If this is not the case, then a careful investigation of the above factors, particularly x-ray wave forms, switching times and sensor positions, should be carried out. It should be possible to select conditions under which consistent results can be achieved.

SYSTEMS CURRENTLY USED IN CHEST RADIOLOGY

Table 2 shows the components of the imaging systems currently in use for chest radiography in both Ontario and Manitoba. It also indicates the trends in the changes that are taking place in imaging systems. These changes are frequent and widespread. For this reason, the number of rooms investigated in detail in Ontario was kept small, because extensive surveys would be out of date within a few months.

Table 2. Components of the imaging systems currently used for chest radiology

Screens	mostly DuPont Hi Plus, Hi Speed or SP or Kodak Regular; a few par speed
Film speed	mostly of speed similar to Kodak RP or DuPont Cronex 4, 6, or 6 plus; some films of half this speed; films may be underdeveloped in order to reduce contrast and increase latitude, and may therefore have a lower speed than for maximum development
Film Contrast	mostly contrast films (Cronex 4, Kodak RP) but a trend toward longer latitude films or underdevelopment
Grids	mostly 5:1 ratio, 80 lines/in; trend to 12:1 ratio, 100 lines/in; trend to high kVp - air gap technique with grid removed
kVp	half the rooms 80-100 kVp and the other half 100-140 kVp; trend toward higher kVp

DISCUSSION

Because the use of photofluorography for chest radiography has decreased to a large extent, the x-ray exposure of patients due to chest radiography has now reached quite low levels. Compared with the exposures due to abdominal examinations, which may reach 100 R in some cases, the chest exposures of a few tens to a few hundreds of milliroentgens are relatively insignificant (4). However, if it is possible to decrease the range of exposures from the current factor of 10 down to about a factor of 2 and at minimal cost, it seems worthwhile to do so. Further reduction, of course, is always encouraged but a much higher priority should be placed on the reduction of exposures from abdominal radiography.

The hazards to be expected from chest radiography are probably leukemia and breast cancer. Indeed, the dominant hazard may be cancer of the right breast in women. This receives the highest dose when both PA and lateral films are taken, up to 200 mR per lateral radiograph in this study and the breast is often completely included in the x-ray field. The hazard of leukemia would presumably be about the same in men and women but only about 20 percent of the total bone marrow will be irradiated and exposure will be even lower than to the breast. However, accurate assessments of the hazard are not possible and therefore, although the hazards are likely to be small, it is still worthwhile to try to reduce the doses from all examinations down to the lowest values found to be acceptable for a satisfactory radiograph.

This study suggests the key factors in this dose reduction. Table 3 lists the components of a proposed technique for achieving low exposures in chest radiography.

Table 3. Proposed techniques for low exposure chest radiography

1. 120 to 140 kVp with air gap, no grid
2. high speed screen (Du Pont Hi Plus or equivalent)
3. film speed equivalent to Du Pont Cronex 4 or Kodak RP
4. film developed to full speed
5. latitude film or latitude development
6. 300 mA generator (forced extinction with higher mA)
7. exposure time 10 to 50 ms
8. properly adjusted phototiming

best copy available

Using this technique, it should be possible to obtain satisfactory radiographs at an exposure of 5 to 10 mR for a PA view. If a grid is used in place of an air gap technique, the exposures will be between 10 and 20 mR, still acceptably low.

While it is reasonable to try to use even lower exposure using faster rare-earth screens, care should be taken not to affect the quality of the radiographic image and the accuracy of diagnosis because this might be a greater hazard to the patient than the danger of cancer (5) Greater priority should be given to the reduction of dose for other x-ray examinations, particularly of the abdomen. Our efforts should now be directed to achieving the reductions of exposure which are simply and readily accomplished.

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PATIENT SELECTION FOR CHEST X-RAYS

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Radiology can be categorized into three distinct but interrelated processes - selection, conduct and interpretation.

Selection is the process by which someone (a physician usually) decides that another person (a patient usually) needs an x-ray examination. The most common situation occurs when a patient's physician requests a radiological consultation. However, there are other people and organizations besides attending physicians that make this decision. For example, an attorney may decide that a client needs an x-ray examination to document physical injury. Congress may decide that compensation for pneumoconiosis will be based upon radiological evaluation. An individual may decide that an x-ray examination is needed to make sure that an ankle is not fractured or that a pneumonia is not present.

The list goes on, but, nevertheless, the process by which the decision is made to have an x-ray examination is defined as "selection." The person that makes the decision is called the "selector." The information upon which the selector bases his decision to request an examination is called selection criteria or referral criteria.

The selection decision triggers the next step, the conduct of the examination. Conduct is the process by which the radiological images are produced. Specifically, the selection of x-ray machine parameters, film, screens, ancillary devices such as grids and gonad shields, the positioning and exposure to the patient, and the development of the images. The "conduct" phase is completed when the decision is made that the examination is adequate and need not be repeated. "Conductors" are usually radiologic technologists but there are others who act as conductors. The conductor initiates the final step of the process by submitting the completed study for interpretation.

Interpretation is the act of drawing inferences about the patient's health based upon the radiological information available. Interpretation is usually performed by radiologists but many other physicians (such as orthopedists, urologists and internists) also act as interpreters.

The three phases are interrelated. The selector can have influence over examination conduct and interpretation. For example, if a selector demands that gonad shields be used for his patients, more than likely they will be used. If the results of an examination are frequently not helpful to the selector, he may discontinue requesting the examination in that situation. On the other hand, if the information is extremely useful, his decision process is reinforced. Likewise, the interpreter has influence over the conduct and selection phases. If the films presented for interpretation are below standard, the conductor will be informed and changes will result. Furthermore, by consulting with the selector about the results of an examination, the interpreter is influencing future selection decisions.

This system can be used to examine the distribution of resources. For example, if the 40 or 50 papers presented at this symposium were classified according to this system, it would be found that all but a few have dealt with conduct and interpretation.

If dose reduction, cost containment and overall efficiency of radiology resources are goals, then there may be productive improvements that could be made in the selection area.

Certainly the examination that is not conducted saves 100 percent of patient exposure and 100 percent of the costs. There can be no greater savings in a single examination than this.

There has been a movement in recent years to commit more resources to the selection area. Increasing numbers of papers are appearing in the literature on such subjects as efficacy studies, referral criteria, high yield criteria, decision tree analysis, and algorithms. All of these can be classified as "selection" activities.

Following hearings before his committee last summer, Congressman Paul Rogers recommended that a national conference be held to bring the sectors of medicine, law, insurance and government together to talk about the efficient use of radiology. The National Conference on Referral Criteria was convened by the Food and Drug Administration's Bureau of Radiological Health last fall in response to that recommendation and the Conference brought together representatives of over 60 organizations. Each person who attended was assigned to one of four workshops which developed detailed reports during the meeting. The reports, along with the proceedings of the meeting, will be available soon; the major outcomes were as follows:

1. The medical professions agreed to work with government to develop referral criteria.

2. Government agreed to play the role of facilitator.

3. The American College of Radiology (ACR) was requested to act as coordinator for organized medicine in securing the appropriate review and evaluation of suggested referral criteria.

4. The concept of using small expert panels to develop referral criteria was endorsed.

Using the outcomes of the Conference as a point of departure, a referral criteria program is being shaped that might operate in the following way:

1. Referral criteria would be prepared by physicians with government acting as the facilitator. This could mean funding studies and research, arranging national consensus meetings, or arranging for small groups of experts to meet as endorsed. There are a number of ways that can be used to develop criteria.

2. The draft criteria would then be transmitted to the ACR which would see that an appropriate review is conducted by organized medicine. The ACR would be relied upon to decide which professional organizations should be asked to review the criteria.

3. Once agreement is reached among the professions, a program of gradual implementation would be followed. That is, the criteria would be tested in perhaps 10 volunteer facilities and then evaluated. Then the testing would be expanded to perhaps 100 facilities, and so forth.

4. At some point, the criteria would be endorsed by the professional organizations, publicized and distributed.

This proposed system is currently in a very open posture and still evolving. Part of our role at UCSF is to establish and manage several expert panels and assist in the development of referral criteria for specific examinations. So far, we have established panels to consider criteria for the utilization of the pelvimetry and pediatric voiding cystourethrography (VCU) examinations and a panel to consider the chest examination. The pelvimetry panel has developed criteria that *are currently under review* by the ACR and the American College of Obstetricians and Gynecologists (ACOG). The VCU panel has developed criteria stating when the examination should be used and should not be used, and a list of situations that should receive further study. The chest panel held its first meeting just last week.

Pelvimetry and VCU examinations are examples of areas of study that have been prompted by an appreciation of radiation risks. For example, pelvimetry has accounted for more than 80 percent of the fetal exposure in this country and studies have shown that it is not particularly helpful in deciding when a cesarean section is necessary. VCU is a high gonad-dose procedure. However, the chest examination is not a high dose procedure. Interest in this examination is sparked because of the high volume (some 60 million annually) and because of cost considerations.

MEMBERSHIP

The chest panel is composed of ten members representing four specialties: industrial medicine, internal medicine, public health, and radiology. Of the six radiologists, three are specialists in chest radiology. One of the public health representatives is a specialist in tuberculosis detection and respiratory diseases and the other is a specialist in radiation protection. In addition to being chief of general medicine, the internist has an MBA and a special interest in medical economics and cost containment. Each member is distinguised and without question an expert in his respective field.

CHEST PANEL

Member	Speciality
Norman Ashenburg, M.D.	Industrial Medicine
Reynold F. Brown, M.D.	Radiologist
John A. Campbell, M.D.	Radiologist
John M. Eisenberg, M.D.	Internal Medicine
David Glasser, M.D.	Public Health
Reginald Greene, M.D.	Radiologist
Russell H. Morgan, M.D.	Radiologist
Gerald S. Parker, P.E.	Public Health
Carl Ravin, M.D.	Radiologist
Nicholas Sargent, M.D.	Radiologist

PURPOSE OF MEETING

The purpose of this first meeting was primarily that of organization, but two main questions were posed to the panel. First, are there categories of chest examinations for which improved referral criteria are needed? We had suggested that government required regulations be considered as a first possibility. Second, do we have proper representation on the panel to undertake this assignment?

RESULTS

Five "selection" areas were identified by the panel as worthwhile for investigation:

1. Occupational case finding for tuberculosis screening;

2. Routine prenatal chest x rays;

3. Routine hospital admission exams;

4. Presurgical or preanesthesia exams; and
5. Periodic chest x rays (annual check-up).

Concerning occupational case finding for tuberculosis detection, we had asked the Conference of State Radiation Control Program Directors to inventory the State and local requirements for chest examinations. Ther results were inconclusive, but nevertheless, they did find in their survey that there was a State or local regulation requiring a periodic chest x ray for more than 20 occupations.

These five areas all represent examinations of asymptomatic people. Much is known about the volume and distribution of chest x rays, such as how many are performed, where they are conducted, the age and sex of the patient, and other demographic data. Why the examinations are conducted is not well-known. The panel suggested that more than 30 million annual examinations are probably performed on asymptomatic people.

The panel also identified three other areas, besides examinations of asymptomatic people, that it wished to investigate:

1. Multiple determinations - This includes examinations of hospital in-patients that have been repeated within a 24-hour period. There is a spectrum of reasons for these multiple exams that range from simple clerical and booking errors to more complicated problems.
2. Portable examinations - There is a suspected overuse of portable chest examinations both for hospital in-patients and for nursing home patients. Because of the lower quality of the examination, the increased cost and the apparent proliferation, perhaps referral guidance may be needed.
3. Quality assurance and repeat examinations in private offices - Most of the government-developed programs and projects directed toward quality assurance have been geared to the large institutions and there is a suspected need to translate these programs into terms applicable to private medical offices.

Each panel member has accepted responsiblity to prepare the draft criteria for at least one of these categories. Each member is free to enlist the support of other colleagues as required and we will try to make available through some means the resources and staff that may be needed.

Concerning membership, it was suggested that a pulmonary physician or cardiologist should be added to the panel.

In conclusion, I would suggest that the reasons for requesting a chest examination not be overlooked as an important part of the process to optimize chest radiology.

GOVERNMENT INTEREST IN CHEST RADIOGRAPHY

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GOVERNMENT INTEREST IN CHEST RADIOGRAPHY

I will try to briefly answer four questions:

1. Why is the government interested in chest radiography?
2. What are the areas of government interest and activity?
3. What government actions have been taken?
4. What government actions can be expected in the future?

Traditionally the practice of medicine has been a very personal affair between the patient and the physician of his (or her) choice. The selection of diagnostic procedures, including chest x rays, was entirely within the province of the attending physician who had seen and examined "his" patient and who ordered diagnostic tests as he saw fit. The only restraint was the patient's ability to pay, and that factor was usually taken into account by the physician. The chest radiographic examination was conducted in accordance with the training of the physician (or the referral radiologist), and the reading or report was used solely to assist the attending physician with patient management decisions. Government involvement in the process was limited to radiation protection standards for x-ray equipment, enforced by occasional inspections.

What has changed? There has been a continuing trend toward group practice and institutional health care. Physician extenders and other paraprofessional personnel increasingly have been interposed between the physician and the patient.

Institutional rules, regulations, and standing orders--designed to improved efficiency and avoid mistakes--have led to routinization of many procedures, including the ordering and conduct of chest x-ray examinations. A "defensive" posture has been assumed by institutions and professionals, not only to avoid legal liability, but equally to avoid omissions, to avoid peer criticism, and to respond to a public expectation of perfection manifested by pressure to diagnose or exclude all possible disease processes regardless of cost, probability of significant yield, or alteration in health outcomes.

The mechanism for payment of health care costs has shifted from direct private to third-party payments, thereby eliminating the major deterrent to the ordering of x rays. With the assumption of the responsibility for payment by third parties has gone the authority or "right" to critically assess both the need for such procedures and, much as the clinician might disapprove, their conduct. Government, through one or another program, now provides over 42 percent of total health care costs (1). As the major payer, government will scrutinize the product for which it pays. That scrutiny will extend to the quality of the product as well as to its quantity.

Increasing public concern for occupational and environmental problems and wider use of public health screening programs add substantially to the use of chest x rays and the resultant burden of radiation exposure and cost. The total impact is not inconsequential. It is estimated that over 75 million chest x rays are being taken annually at a total cost close to

$2 billion, resulting in radiation exposure to the chest estimated at 2 million R (2). Considering the intensity of national interest and attention focused on the recent incident at Three Mile Island where preliminary estimates of human exposure were 100 millirem individual maximum and 3,000 to 4,000 man-rem total body, it does not seem inappropriate for the government to express more than a casual interest in chest radiography, even though the exposure to any individual from a single chest x ray seems trivial (3).

What are the areas of government interest in chest radiography? They can be stated simply: Cost and radiation exposure. The goals are equally simple and they coincide: Reduction! Reduction in cost can be achieved most easily by restricting or reducing unnecessary examinations. Other, more subtle, cost reductions can be achieved through improved technical, interpretive, and administrative procedures. Reduction in radiation exposure can be achieved in numerous ways, many of which are being addressed at this conference. I believe the perspective of the Bureau of Radiological Health coincides with that of most of this audience. Namely, that reduction in radiation should be sought where, and to the extent that, it will not compromise or reduce the useful and usable information in the image. Therefore, the stated and actual goal is to encourage technological improvement and innovation that will increase or enhance information yield per unit of absorbed radiation in diagnostic imaging.

What government actions have been taken? Several specific actions impact directly upon the selection of patients for, and the conduct of, chest x-ray examinations.

With reference to exposure reduction, there is a Federal performance standard for diagnostic x-ray equipment (4). It applies to all such equipment manufactured after August 1, 1974. The requirements are imposed on the manufacturer and the assembler and have resulted in more reliable exposure factors, better shielding, and beam definition or limitation, all contributing to reduced patient and operator exposure. A Federal recommendation on the use of gonad shielding has received wide notice and acceptance. Other recommendations, such as those on quality assurance and intensifying screens have been proposed and ultimately may be formalized. These are recommendations and are intended as advice or guidance. They are not legal requirements. Similar guidance may result from conferences such as this in the form of either recommendations or technical reports.

The Federal government has issued no regulatory standards for, or limits on, radiation exposure from the use of x rays for diagnostic imaging. In February of 1978, the President did issue a directive titled "Radiation Protection Guidance to Federal Agencies for Diagnostic X rays" (5). That Guidance approved and promulgated recommendations issued jointly by the Adminstrator, Environmental Protection Agency, and the Assistant Secretary for Health, Department of Health, Education, and Welfare, under their respective authorities and mandates. It contained recommendations "made for the guidance of Federal agencies in their conduct of radiation protection for diagnostic uses of x rays in the healing arts." Recommendation No.10 addresses the exposure from diagnostic x rays and is reproduced below:

> 10. Technique appropriate to the equipment and materials available should be used to maintain exposure as low as is reasonably achievable without loss of requisite diagnostic information; measures should be undertaken to evauate and reduce, where practicable, exposures for routine nonspecialty examinations which exceed the following Entrance Skin Exposure Guides (ESEG):

Examination (Projection)	ESEG (mR)*
Chest (P/A)	30
Skull (lateral)	300
Abdomen (A/P)	750
Cervical Spine (A/P)	250
Thoracic Spine (A/P)	900
Full Spine (A/P)	300
Lumbo-Sacral Spine (A/P)	1000
Retrograde Pyelogram (A/P)	900
Feet (D/P)	270
Dental (Bitewing or Periapical)	700

*Entrance skin exposure determined by the Nationwide Evaluation of X-Ray Trends program for a patient having the following body part/thickness: head/15 cm, neck/13 cm, thorax/23 cm, abdomen/23 cm, and foot/8 cm

Although not directly applicable to the private sector, this Federal Guidance will impact broadly on all Federal agencies and activities and has implications beyond Federal facilities. It established specific exposure levels that are intended as guidelines or maximum target levels for Federal facilities.

The entrance skin exposure for PA chest examinations is given as 30 mR. That figure was derived from the results of a nationwide survey referred to as NEXT and described elsewhere in this Conference. It approximates the third quartile of actual measured exposures and was intended as a goal or target level. Recognizing the broad range of exposures associated with what are considered acceptable chest techniques, the committee charged with implementing the Guidance is reassessing the value for chest (P/A) and may recommend a change.

There is precedent for guidelines or regulations of this nature. In 1974, Illinois established a regulation containing similar guidelines (6). That regulation is a legal requirement; similar regulatory action has been taken by Vermont and Minnesota and is proposed in other States. If a disparate or inconsistent variety of such regulations should ensue in other States, it may become desirable for the Federal government to intervene to establish consistency or standardization among the various States. The major mechanism for encouraging consistency in State regulation in this area is the Suggested State Regulations for Control of Radiation (SSRCR). The SSRCR, first published in 1962, is under continuous review and updating. It is a joint effort of the Conference of Radiation Control Program Directors, Inc., the U.S. Nuclear Regulatory Commission, the U.S. Environmental Protection Agency, and the U.S. Department of Health, Education, and Welfare's Bureau of Radiological Health (BRH). The project is coordinated by the BRH.

The Bureau of Radiological Health is considering an exposure guideline proposal that would be oriented toward commonly used and acceptable techniques for common procedures. Chest radiography would be included in such a guideline or recommendation.

A more formally structured and detailed set of "recommendations" is included in the Specifications for Medical Examinations of Underground Coal Miners issued by the National Institute of Occupational Safety and Health as part of its responsibility and authority under the Federal Coal Mine Health and Safety Act of 1969 as amended (7). The Act requires chest x rays to evaluate pulmonary disease in coal miners. It also addresses in some detail the conduct of the chest examination and includes sections on film size, positioning, focal spot size, generator capacity, exposure time, film and screen speed, processing and quality assurance. The specifications are not enforced upon the private sector, but by virtue of their character as "conditions of participation" in the program of coal miner chest examination, x-ray facilities located in coal mining areas would find it awkward to decline to participate in a program critical to the health and welfare of a substantial fraction of their constituent medical population.

What government actions relate to the selection of patients for chest radiography?

In 1958 the Public Health Service issued an official policy statement on the use of mass radiography in tuberculosis control. It recommended, among other things, that x-ray case-finding be applied selectively in groups at high risk of chest disease and that every community evaluate its tuberculosis problem, need, and resources so that local chest disease surveys could be maximally efficient and effective.

The 1958 statement was superseded on February 18, 1972, when the Assistant Secretary for Health and Scientific Affairs signed a Department of Health, Education, and Welfare policy statement on the use of the chest x ray as a screening procedure for cardiopulmonary disease (8). That statement makes two principal recommendations:

> The community chest x-ray survey among the general population should not be used as a screening procedure for the detection of tuberculosis, other pulmonary disorders, and heart disease.
>
> When chest x-ray screening programs are justified for selected population groups, the full size radiograph, rather than the miniature film used in photofluorographic equipment, is preferred.

The Presidential Guidance of February 1978, referred to above, also includes sections or recommendations addressing the question of ordering chest radiographs. Relevant sections are reproduced below:

> 2. Prescription of x-ray studies should be for the purpose of obtaining diagnostic information, should be based on clinical evaluation of symptomatic patients, and should state the diagnostic objective and detail relevant medical history.
>
> 3. Routine or screening examinations, in which no prior clinical evaluation of the patient is made, should not be performed unless exception has been made for specified groups of people on the basis of a careful consideration of the magnitude and medical benefit of the diagnostic yield, radiation risk, and economic and social factors. Examples of examinations that should not be routinely performed unless such exception is made are:
>
> a. chest and lower back x-ray examinations in routine physical examinations or as a routine requirement for employment,
>
> b. tuberculosis screening by chest radiography,
>
> c. chest x rays for routine hospital admission of patients under age 20 or lateral chest x rays for patients under age 40 unless a clinical indication of chest disease exists,
>
> d. chest radiography in routine prenatal care.

This is intended as guidance for Federal agencies and is primarily for Federal facilities. It addresses the ordering of x rays. For x-ray screening it advises that there should be deliberation, with a consideration of the indications for screening prodedures, and their institution only after evaluation of risk and cost factors and a determination that anticipated benefit will justify them. This same advice has been circulated to the private sector for information, but not as a formal recommendation (9).

There is another source of "recommendations" to the medical community from the government. It is in the form of conclusions and implied recommendations from the National Institutes of Health, Consensus Development Conference on Mass Screening for Lung Cancer, held in Reston, Virginia, on September 18-20, 1978 (10). It was sponsored by the National Cancer Institute, in collaboration with the National Institute for Occupational Safety and Health. Relevant extracts from those conclusions include the following:

1. Current studies (screening of asymptomatic individuals) do not at present show any evidence of significant reduction in mortality from cancer.

2. Until their value is demonstrated, mass screening programs should be limited to controlled studies.

6. The control of quality in performing and interpreting chest radiography is essential. Radiographic quality can be assured through supervision by qualified radiologists. Accuracy of interpretation in screening can be enhanced through double reading of chest x rays by appropriately trained individuals.

Another forum that critically examined the selection of patients for radiography was the National Conference on Referral Criteria for X-Ray Examinations held in October 1978 in Washington, D.C. That Conference was suggested by Congressman Paul Rogers and was cosponsored by Congressman Rogers and Secretary of Health, Education, and Welfare Joseph Califano (11). Among the recommendations emerging from the Conference was the suggestion that development of x-ray referral criteria take into consideration cost, as well as the conventional medical risk/benefit questions.

What government actions can be expected in the future? It is certainly reasonable to anticipate more of the same. Where areas of clear public health benefit (i.e., earlier, more precise diagnoses or reduction of adverse effect) appear reasonably achievable, government agencies will initially suggest, and eventually insist on, their adoption.

Where there is apparent overutilization or demonstration of ineffectivness of procedures, the government will initially suggest and eventually require that clear indications and/or justification be recorded. That may be accomplished by regulations or by recommendations, but clearly the most effective tool is the pursestring. For that part of health care payments that is within the province of the Health Care Financing Administration, the signals are clear. The Government will not continue to pay for procedures that are not both medically appropriate and necessary. The Congress has spoken on that issue in the Medicare Amendments to the Social Security Law and the Administration will execute that mandate.

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DISCUSSION

DR. RAO: One of the problems in patient selection seems to be that radiologists have become much more technology oriented than other specialists. They do not ordinarily see their patients. The examinations are ordered by clinicians. For this reason, the radiologist lacks influence on the selection of patients and on the proper sequence of conducting the examinations. That is the main problem as far as reducing exposure is concerned.

DR. JOHNSON: Let me just comment that I think we have a tendency to pursue the path of least resistance. Changing the attending physicians' referral practice is not a simple task.

DR. FIELDS: I was a bit amused, as I think all of us were, by the recent two steps forward and three steps backward taken by the FDA and BRH on the extension of the 1974 regulation that they have given us as a basis for evaluating cost-risk as $100 per man rad per year (Shope, T.B. and H. Rudolph. A second look at costs and benefits associated with the diagnostic x-ray equipment performance standard and policy on assembly and reassembly, Draft #3, Bureau of Radiological Health, Rockville, Maryland, November 1978). Relative to the State of Illinois limits of exposure, with which I am familiar, I have figured that it cost the State of Illinois $30 million dollars minimum for the sales efforts relating to high-speed screens and everything else required to meet those exposure limits. The actual benefit per dollar in Illinois should be examined before we impose national limits on exposures.

DR. JOHNSON: I wish we had a little more time to go into that. I might ask if you would suggest another figure, such as $200? The $100 figure that was quoted was not used by the Bureau of Radiological Health for any public health purpose. It had a unique purpose, explained in the document to which you referred. We do not have time to discuss this adequately, but please do not be left with the wrong impression.

DR. GREGG: I would just like to state that the only economical way government regulations can be enforced is by documentation--the accumulation of paper. Yet no one ever takes into account the enormous number of man-hours that are spent just in documenting something that you may or may not have done depending on your degree of honesty. I hope that, if nothing else, out of my talk on radiation risks you will perceive that the regulation of 30 mR per exposure is ridiculous and an exercise in futility. When you look at the enormous ranges of our uncertainty concerning radiation carcinogenesis, and the very low probability of occurrence for any imaginable sequence of chest films, does it make any difference whether you are limited to 30 mR or 60 mR? We have no proof that this regulation is going to accomplish anything except possibly lower diagnostic accuracy. Since we cannot conceivably measure the dose under actual clinical circumstances, I feel that its documentation is truly an unnecessary expense.

DR. LOGAN: I just wanted to make a quick comment on the question of a lateral chest film. Many women in the United States have a lateral chest film every year. While I believe it is indicated in many instances, we ought to think twice before we do a routine lateral view on women under age 30, because the breast x-ray dose is much higher on the lateral view than on the PA. We are looking into the fact that the left breast is, by convention, placed close to the film on the lateral view with the right closer to the x-ray source. The incidence in the left breast has been 1.1 times that in the right beginning in 1946, the first year for which records are available (Haagensen, C.D. Diseases of the Breast. W.B. Saunders, Philadelphia, Penna., 1956). Since the right breast is getting more dosage than the left, we are trying to determine whether breast cancer incidence is skewed more to the left breast. I doubt whether the small x-ray dose from chest films will influence this much. However, in young women a lateral chest film is seldom necessary and I would not recommend it as a routine. That same dose could be better used to diagnose breast carcinoma by mammography (Logan, W.W. and E.P. Muntz. Reduced Dose Mammography. Masson Pub., New York, NY, pp. 407-414, 1978).

DR. NICKOLOFF: I think the NEXT program is worthwhile and is widely quoted, but I think there are several things of which people should be aware. There is a large variation in exposure with variations in technical factors. In fact, you can have at least a 100-fold change in exposure with changes in such factors. In film/screen combination, kV, filtration, and distance. A single number quoted for exposure may be much too high for someone using a "fast" system, and too low for a person using a "slow" system. Another interesting point is that the NEXT data were collected by surveyors in many States (not measured on patients) with different kinds of equipment. The surveyors asked people what technique they were using for the standard size patient. I think this survey methodology has an influence on the results. Also, as we saw in the graphs Dr. Properzio showed us, the high kVp and phototimed systems were not included. I think these biases in the study have a great impact on the final result. Finally, if one were to look at the NIOSH Study and the pneumoconiosis panel recommendations, he would conclude that a par speed system should be used at high kV with high grid ratios. I think we should examine these recommendations carefully. If you follow all of them for a 24-to-25-cm chest, your average exposure for a PA chest would be around 50 mR, which would exceed the guidelines.

DR. JOHNSON: May I give you a free plug? Dr. Nickoloff was kind enough to let me see a preprint of an excellent paper he submitted to RADIOLOGY. If my recollection is correct, the experience he reported there for a 23-cm chest was somewhere close to the NEXT average, i.e., around 23 to 25 mR, with Hi-Plus screens. We just want to commend you on being under the President's guidelines.

DR. PROPERZIO: I will not respond directly to the comments on the NEXT program. John Cameron and John Wochos have published some information on NEXT pointing out some of its shortcomings. I would like to make a personal comment with respect to the exposure guides--30 mR, or whatever the number might be. I think people get overly concerned about that number. Dr. Johnson pointed out that the recommendation in the Presidential Guidance says only that you must consider the stated Entrance Skin Exposure Guides, not that your exposures must be below them. I think the essence of what we should be striving to do is to have a facility analyze the way they are doing their chests and other procedures. We have not really looked at a massive number of radiographs in chest radiography. In dental radiography and mammography, we have. Our BENT Study has shown that a great majority of facilities doing mammography do not know what exposure they are delivering to patients, and a high percentage are doing something wrong. Their processing is not optimal or they have mismatched the energy spectrum to the image receptor. A number, such as that provided in the exposure guides, prompts a facility to evaluate its operation and compare itself against reference values. Where problems obviously exist, corrections can be made. If we can bill it simply as a recommendation, we can point to the fact that a facility should know what it is doing--not just be able to tell you "Oh yes, we have a tube with a certain focal spot; we have a Kodak processor, and we have a certain film/screen combination." Knowing what equipment you use does not necessarily optimize your imaging system. The exposure guide number helps a facility compare itself with what has been reported for a large number of facilities. I think that it is the more important point as opposed to an undue concern with forcing everyone to achieve a certain number.

DR. STITIK: Dr. Nickoloff and I work together on the lung cancer screening project. At one point in the project, we were required to switch to high-speed screens and films. I do not like them! Subsequent to the change, some subjects who have been in the project for 4 years now have nipples and rib crossings showing up for the first time. We are recalling a large number of these patients for repeat films with nipple markers, oblique views and tomography. In my opinion, this has led to an unnecessary radiation exposure of these volunteers as well as causing them unnecessary concern.

Also, the discrepancy that was read into my presentation really does not exist. I helped write the Reston recommendations (NIH Consensus Development Conferences: Summaries, Vol. I National Institutes of Health, Bethesda, Maryland, 1977-78. Summarized in JAMA 241:1582, 1979.) In my presentation today I clearly indicated that these are preliminary observations and that we are basing our project, not on resectability rates, but on a reduction of death rates from lung cancer. At present, there is no significant improvement in overall lung cancer mortality produced by screening in the three Centers. However, the period of observation has been brief, and a long-term benefit is expected.

FORCED EXPIRATORY TIME RADIOGRAPHY

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The presence of air trapping within the thorax may be an important clue leading to the detection of occult pulmonary disease. Partial or complete bronchial obstruction, or destruction of the supporting interstitial tissues which serve to keep the bronchi open, may cause trapping of air distal to the zone of obstruction. In partial bronchial obstruction, air may be able to be inhaled past the zone of obstruction on inspiration, but on expiration a flap or ball valve type of action may prevent the egress of air. In complete obstruction, trapping is caused by collateral air drift from non-obstructed areas through interalveolar and bronchiolo-alveolar connections. This keeps the lung distal to the obstruction inflated. Since mechanically it is easier for air to collaterally drift into an area of obstruction than for it to get out, trapping occurs.

Oxygen is absorbed from the trapped air. The residual hypoxic gas mixture causes vascular spasm and blood is diverted away from the area of trapping to normal lung tissue. On the radiograph, this produces radiolucency (oligemia). When the trapping is marked or long-standing, such changes may be visible on the standard inspiratory roentgenogram. However, in many instances, the inspiratory roentgenogram fails to indicate conclusively or fails even to suggest the fact that trapping is present. A roentgenogram taken in expiration will usually detect the presence of trapping. The radiologic signs indicating trapping are well known and consist of failure of motion of the diaphragm on the side of the trapping, mediastinal shift to the opposite side, failure of the rib interpaces to narrow on the side of the trapping, and the presence of oligemia in the trapped zones. However, given sufficient time to expire forcefully, an individual with areas of trapping in his lung may be able to get rid of that air on expiration and the trapping will go undetected. This phenomenon is true for both localized and generalized trapping, making the radiologic diagnosis of obstructive generalized airways disease a difficult one, and only reliable in the late phases of the disease.

We have adapted our expiration radiographic series to include a film exposed 1 second after the patient commences the forced expiratory maneuver. A commercially-available timing vitalometer which produces an electric signal 1 second after expiration commences is hooked into the generator. The patient practices the forced expiratory maneuver with the mouthpiece of the vitalometer in place. An inspiratory film is then made, again with the mouthpiece in place. The patient is then instructed to begin the forced expiratory maneuver. When he does this, he automatically activates the vitalometer signal and 1 second later, during expiration, an x ray is obtained. A full expiration film is also frequently added to the series.

Utilizing this technique, we have been able to detect minor degrees of localized air trapping in the presence of normal standard inspiration and expiration films. We have also been able to say with certainty that patients have generalized airways obstruction long before any morphologic change of obstructive pulmonary disease is visualized on the roentgenogram. We have measured lung volumes from these roentgenograms and in the normal these correlate very well with volumes determined utilizing planimetric methods or utilizing a series of simple linear measurements.

The technique is simple but does require a radiographer to be trained in demonstrating to patients the method of performing a forced expiration. In addition, several practice maneuvers with recording of the movement of air on the vitalometer are essential, since if the patient does not produce maximum effort on the forced expiratory maneuver when the films are being made, the record on the vitalometer will differ from those of the practice maneuver, and the roentgenograms may indicate that generalized trapping is present (false positive for obstruction airway disease). We have seen no studies in which trapping has been demonstrated on the full expiration film and not seen on the film made 1 second after commencing expiration. We have several instances in which no trapping was seen on the full expiration film, but it was clearly demonstrable on the film made at FEV_1 (Forced Expiratory Volume in 1 second). Studies are currently underway to determine whether the lung volumes obtained from the roentgenogram and from plethysmography have a high correlation in patients with localized and generalized obstructive pulmonary disease. In addition, we have a modest series of patients who have had roentgenograms utilizing the FEV_1 technique and have also had xenon inhalation studies. These are currently being compared as well.

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QUESTIONS

DR. CAMERON: May I ask what your exposures are, typically.

DR. GREENSPAN: Our exposure durations run from 6 up to around 30 milliseconds—that is for the lateral in a very heavy patient, and utilizing the Lanex and Ortho-G systems. We use the 0.3 mm focal spot in small patients.

Dr. Ravin and I have spent a little time trying to figure out which would be more advantageous: to use the 0.3 mm focal spot at low mA with a slightly larger exposure or the shorter exposure time but greater geometric unsharpness with the 0.9 mm focal spot. We could not decide. We could not see any detectable difference and I felt quite strongly that this experiment should be done on patients not on phantoms because phantoms do not move and patients do.

We figure that with this technique we are well within any proposed new Federal regulations on dosage.

DR. CHARBONNIER: I would assume that there would be some motion of the patient during the process of forced expiration. Does this pose any problem in the comparison of films? Is there any motion blur during the second film?

DR. GREENSPAN: Not appreciably. Of course, when we initially set the technique up, we were very worried about this being the case, but we were quite amazed that we do not see significant blurring. I would not want to use that second film for the most detailed analysis of distribution of small pulmonary vessels. I still do not see any motion on it but I am afraid there has to be some.

I might say that it would easily be possible to have an exposure time of less than 6 milliseconds, but we are limited by a generator that does not have forced extinction. We are using a rather old Ionomat and when we get down to around 8 milliseconds it is not reliable. We will be switching to a forced extinction generator and, if we use slightly higher mA rating, we will be able to get films made at a shorter exposure duration. The disadvantage of that, we feel, will be that we will have a bit more "bloom" on the focal spot.

MR. FIELDS: Have you found any particular advantage to the inspiration and expiration chest radiographs over xenon studies for determining lung volumes?

DR. GREENSPAN: While I think that the two studies show the same thing, we have a couple of advantages. Number one, we can show morphology at the same time. With the xenon, you are looking at gas retention but you do not have the chest detailed anatomy to look at. In other words, we can look at a lesion or multiple lesions at the same time. The other advantage is that radiographs can be performed much more rapidly than a xenon study. Dr. Gottschalk and I are comparing the two studies in a group of patients right now. I would certainly expect that the xenon method will turn out to be somewhat more sensitive, but we have not seen any gross discrepancies. The xenon study has one advantage over the FEV_1 study: you do not need as cooperative a patient. You can do xenon studies in much sicker patients than the FEV_1 study.

PORTABLE CHEST RADIOGRAPHY

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The mention of portable chest radiography usually evokes the response that these are the poorest quality examinations performed on the most seriously ill patients. Films taken at the bedside comprise an important segment of inpatient chest radiography in any large hospital but particularly in secondary or tertiary care centers. In 1978 at the University of Wisconsin Clinical Science Center, 41.6 percent of all inpatient chest films were taken at the bedside. Beyond sheer numbers, bedside examinations assume importance because they usually involve the seriously ill patients. Unfortunately, we must accept as fact that films taken at the bedside will always be inferior to first-rate films taken with stationary equipment in the Radiology Department but certainly the best mobile equipment today produces films superior to those produced by standard single-phase units of just two decades ago and with those images radiologists were making some brilliant diagnoses (or at least they thought they were).

With seriously ill patients in the intensive care unit (ICU) or surgical recovery, the risk and difficulty of transportaiton to the Radiology Department outweighs the benefit of improved films, and this is the logical basis for ordering bedside radiographs (1). Some institutions are now incorporating stationary overhead tubes and generators into the ICU at some considerable expense and decreased flexibility. Before any expensive modifications of practice or equipment are undertaken, the question must be answered whether substantial, if any, improvement in patient care or even advancement in radiodiagnosis will result.

What are the purposes of portable chest radiography and how are they advanced by proposed improvements in radiographic technique? And at what cost? What aspects most need improvement? To what extent are the limitations purely technical and to what extent are those limitations imposed by the patient's physical condition?

I have examples of superb AP bedside examinations which compare favorably with PA examinations performed in the department. In many cases, it would be difficult to pick the standard film over the one taken at the bedside. Available remote technology can produce excellent quality radiographs fully on par with those taken using stationary equipment as long as certain conditions are satisfied. Namely, that the patient is of small or average size and fully cooperative with no impairment of cardio-respiratory function. Obviously, patients who have bedside chest radiographs rarely fulfill these conditions and short of miraculous restoration to health, they tend to yield suboptimal radiographs regardless of equipment employed. These patients are seriously ill, encumbered by various tubes and wires, unable to assume or maintain an optimal erect position for proper filming.

Two AP portable films are taken of the same patient one day apart. On one, the technologist managed to film the patient upright and the moderate pleural effusion (2000 cc) is obvious. On the second, the technologist settled for a recumbent film and the presence of an effusion can only be guessed. Radiologists are well aware of the effect of position changes on the pulmonary vasculature but it is difficult to interpret when the degree of recumbency is either unknown or varies from day to day. Steelshot and mercury drop markers are of value only in documenting that the film was exposed in a horizontal position. They are of little, if any, value in documenting upright position. Anyone who wishes to may convince himself of this by taping such a marker to a fluoroscopic tilting table and observing it as the table is put through its range of motion. Gallant and others have devised an elegant method for documenting various degrees of cassette inclination (2). Lacking such device, I consider the patient was upright only when I see a fluid level within the stomach.

Patient rotation is another common problem with AP portable films. As little as 5 to 10° rotation may cause the mediastinum to appear wide and the cardiac silhouette to appear abnormal. There is also a tendency for portable films to be taken in a somewhat lordotic projection, probably because the apparatus is usually positioned at the foot of the bed. This lordotic projection causes geometric distortion in the cardiac size and foreshortening of the entire chest as well as obscuring the posterior lung bases, an area where fluid usually collects in the supine patient.

Another major problem relates to lack of respiratory control on the part of the patient, who is rarely able to suspend respiration at total lung capacity and is often comatose or otherwise unable to cooperate. A rapid respiratory rate also makes timing of exposure extremely difficult. All these problems result in less than optimal films but relate to the patient's condition. These problems are not likely to be completely resolved by modifications of radiographic equipment or technique.

Another important aspect to consider is the nature of the information expected from bedside chest radiographs. This differs vastly from the routine PA film ordered on an outpatient. A patient presents in the emergency room short of breath and a chest film is ordered, the radiologist may make an astute diagnosis of acute pulmonary edema or pneumonia and become a hero. Things are quite different in the intensive care unit where minute ventilation, A-V O_2-difference, pulmonary artery wedge pressure and possibly even left ventricular ejection fraction and extravascular lung water are continually monitored. The radiologist is usually last to know when the patient lapses into congestive failure. Liebman in a study comparing radiographic versus isotopic estimation of pulmonary edema found a very poor correlation (3,4). Indeed, he found that isotopic estimation of extravascular lung water correlated best with radiographic estimation of pulmonary edema on films taken 24 hours later. With patients being so closely monitored, what information is the bedside chest x ray expected to supply?

I reviewed the course of 82 critically ill or post-surgery patients who were hospitalized at University of Wisconsin Hospitals in November of 1978. Three hundred thirty-two bedside chest examinations were performed on these patients (Table 1). Forty-three percent of these were taken following placement of vascular catheters, electrodes, pleural drainage tubes or endotracheal intubation; 7 percent were taken following other invasive procedures such as bronchoscopy or thoracentesis. These films were ordered not only to document the lack of complications such as pneumothorax, pneumomediastinum or pulmonary hemorrhage but also to verify position of the various devices inserted. Only 24 percent of films were ordered to evaluate pulmonary disease such as pneumonia, aspiration or pulmonary embolism. The 6 percent ordered specifically to exclude congestive heart failure were usually in patients without intravascular monitors, or in whom the clinicians wished to exclude primary lung disease as the cause for abnormal pressure readings. The point to be made here is that catheter and tube localization is certainly an important role for bedside chest radiography as it stands now. Tabriski and associates with extensive experience in remote high kilovoltage technique (240 kVp) have found that certain widely-used polyethylene catheters are poorly imaged by this technique (5). This problem, though easily correctable by suitably radiopaque catheter markers, points up the importance of assessing technical innovations in light of what is the prime clinical or medical concern.

Table 1. Analysis of indications - 332 bedside chest exams

Catheter/tube placement	42.7 %
Following other procedures	7.3
? Pneumonia/aspiration	17
? Pulmonary embolism/infarct	7
? CHF	6

A large group of indications (19.5 percent) for bedside chest radiography is to monitor patient progress—that is, the clearing or worsening of lung disease. In order to do this effectively, films must be comparable in technique from day to day. Anyone who has had to interpret portable chest films is quite familiar with the problem of deciding whether changes in the appearance of lungs are real or due to variation in technique. This is one area of chronic deficiency deserving more attention on the part of radiologists, technicians and equipment manufacturers. The prospect of automated exposure in this area is exciting but not without its potential problems. The very fact that such systems are designed to achieve a general level of film density, while producing an acceptable radiograph, may also obscure real changes in subject density. There are physiological as well as technical factors which may cause variation in the opacity of lung parenchyma from day to day. The depth of inspiration as mentioned previously is difficult to control in a very ill patient and may vary greatly between exams. Siemens has developed a method of triggering x-ray exposure at any given point in the respiratory cycle of those patients on automated ventilator support by their system (Siemens-Elema servo ventilator 900B). The possibility of relating radiographic appearance to any particular level of lung inflation opens many interesting avenues of investigation.

I also analyzed the 332 bedside chest examinations for technical quality (Table 2). This was a "retrospective" analysis in that the studies had been accepted by the radiologist or clinician. Films were taken with General Electric AMX-II battery-powered mobile x-ray unit, Dupont Hi-plus screens and Kodak XRP-1 film. Film-focus distance was from 56" to 72". With bedside films we have experienced an approximate 5 percent retake rate which seems good on the face of it but derives in part from the difficulty in obtaining the examination. Often the attending physician will refuse a retake when it has been suggested and, knowing that the patient is having daily or more frequent examinations, the radiologist may pass a film that would otherwise be rejected. Surprisingly, I found only 10 percent of these films to be marginal or unacceptable. We must assume they all supplied sufficient information for management of the patient at the time they were taken. Interestingly, as many films were over- as under-exposed. If low output of portable equipment were an important problem, one would have anticipated many more under- exposed films. Five of the eleven under-exposed films were of patients postpneumonectomy or with opaque hemithoraces for other reasons such as a large pleural effusion or tumor. The contralateral lung in each case was properly exposed, but it was impossible to assess the dense side of the chest. A wide latitude system would allow imaging of dense portions of the chest (mediastinum, retrocardiac and posterior lung bases, as well as areas obscured by effusion or consolidation) without overexposure of normal lung. In such cases, a high kVp technique or a double cassette (Fig. 1) such as described by Wilkinson and Fraser (6) would be useful in extending the latitude, the latter without reducing subject contrast. But extended latitude may be had without expensive equipment or modification by appropriate choice of screen (7) or film (Fig. 2).

Table 2. Quality analysis - 332 bedside chest exams

Overexposure	11
Underexposure	11
Objectionable scatter	8
Positioning poor	4
Processor fault	1
Marginal quality exams	**35 (10.5 %)**

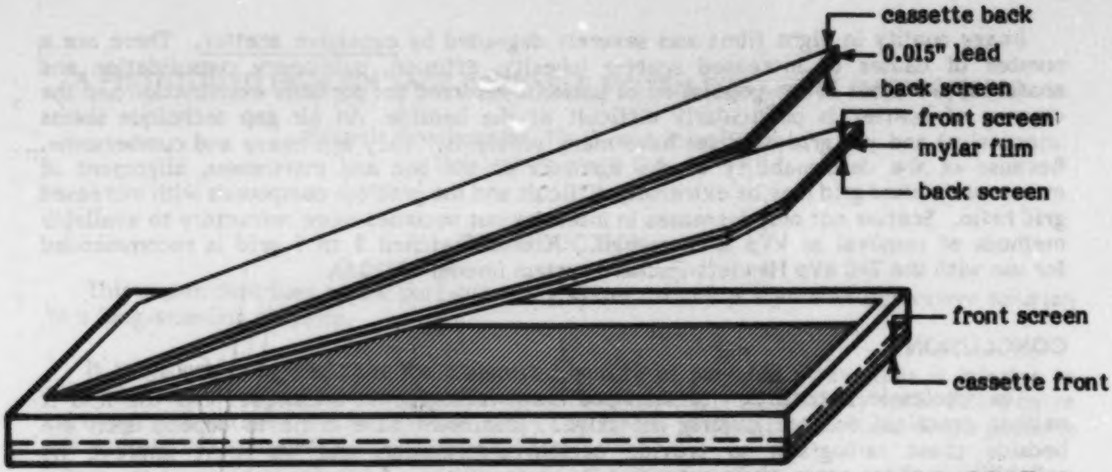
cassette back
0.015" lead
back screen
front screen
mylar film
back screen
front screen
cassette front

Figure 1. Components of a standard three-strap cassette fitted with two pairs of Du Pont Hi-Plus screens. The two middle screens are bonded back-to-back on a thin mylar sheet 3/8 inch (0.95 cm) longer than the screens, whose edges are sealed with thin transparent tape. This combination is hinged at the hinge-end of the cassette; one outer corner is cut diagonally, to facilitate insertion and removal of the front film. Standard felt backing (not shown) is used. (From: Wilkinson and Fraser, Use of a Double Screen-Film Combination in Bedside Chest Roentgenography. Reprinted by permission of The Radiological Society of North America, Inc.)

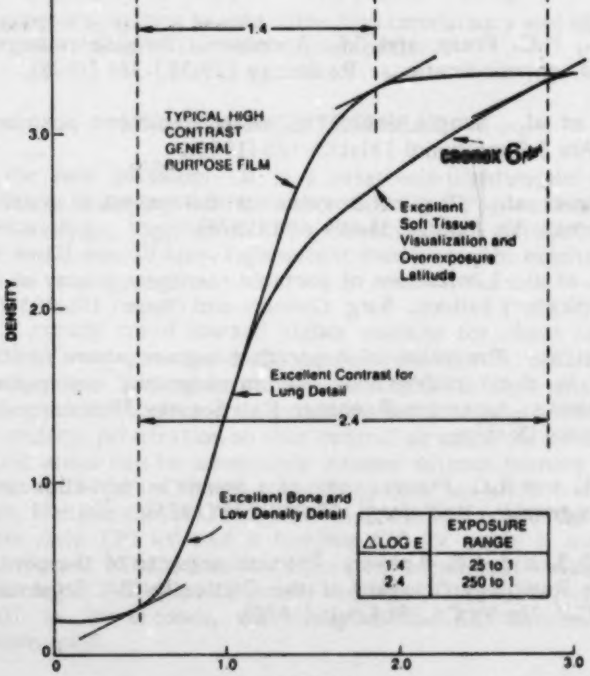

Figure 2. H & D curve of an extended latitude film.

Image quality in eight films was severely degraded by excessive scatter. There are a number of causes of increased scatter (obesity, effusion, pulmonary consolidation and anasarca) prevalent in the population of patients referred for portable examination and the cleanup of scatter is particularly difficult at the bedside. An air gap technique seems impractical and yet grid casettes have many problems. They are heavy and cumbersome. Because of the deformability of the surfaces of the bed and mattresses, alignment of even an unfocused grid may be extremely difficult and the problem compounds with increased grid ratio. Scatter not only increases in intensity but becomes more refractory to available methods of removal as kVp is increased. A cross-hatched 5 to 1 grid is recommended for use with the 240 kVp Hewlett-Packard system (model 43820A).

CONCLUSION

The wholesale importation of standard chest radiographic techniques into the ICU is neither practical nor particularly effective. Clinicians have come to depend upon the bedside chest radiograph to provide certain information and we must improve its reliability in those areas while extending its performance. Adequate imaging of the entire chest, that is the lungs, lung bases, mediastinum, and even the subdiaphragmatic area of the upper abdomen, must be provided. Efforts must be made to improve reproducibility in patient position, depth of inspiration, and rendering of lung density. Improved methods of scatter cleanup must be developed for bedside use.

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A PHOTOTIMED HIGH VOLTAGE PORTABLE SYSTEM FOR CHEST RADIOGRAPHY

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Hewlett-Packard Co.

INTRODUCTION

This paper describes a new portable x ray system which presents an innovative solution to a long-standing problem.

It is generally agreed that the diagnostic quality of portable x ray films is inferior to that of x rays obtained in central radiology. The problem is particularly acute in intensive care units which have a high need for x rays (up to one per bed per day) and where patients are acutely ill so that retakes are not tolerable.

Inputs from a number of radiologists helped to identify the factors adversely affecting the diagnostic quality of portable x ray films in ICUs and to define the equipment design characteristics required to meet the ICU need for high quality, consistent chest films. These characteristics include:

1. Small, lightweight, easily maneuvered system;
2. High output voltage to achieve more uniform penetration of all areas of the chest;
3. Short exposure time to minimize motion blur in often uncooperative patients; and
4. Accurate exposure control to maximize film consistency and allow more effective comparison of successive films on a given patient.

HIGH VOLTAGE TUBEHEAD

Figure 1 shows the new portable. It is a relatively lightweight high voltage field emission system with remote automatic exposure and with microprocessor controls. The use of field emission technology, together with a novel voltage doubler tubehead, made it possible to reconcile small overall size, lightweight maneuverable tubehead and high output voltage.

There has been a steady trend toward higher voltage for chest radiography, with a voltage in the 120-150 kV range now generally preferred in central radiology. Conventional portables are generally more limited in voltage, yet in ICUs high voltage is particularly beneficial because diagnostic requirements generally demand good visibility in all parts of the chest, i.e., more uniform penetration so that central airways, mediastinum, retrocardiac and infradiaphragmatic areas can be adequately imaged without burning out the lung field. The new unit operates at 240 kV. Current technology does not allow transmission of such a high voltage in a small, flexible cable. To overcome this problem the high voltage generator in the cart generates only 120 kV and a flexible 120 kV cable is used to connect the generator to the tubehead. The latter incorporates a compact voltage doubler circuit around the tube producing a 240 kV pulse across the tube. Typical exposure times for a PA chest range from .01 to .04 seconds, depending on the SID and on the speed of the film/screen combination used.

A grid must be used at this high voltage to control scatter. A crosshatch 4:1 ratio focused grid was selected because it reconciles effective scatter cleanup with the wide focal range and tolerance to misalignment which are needed for portable operation.

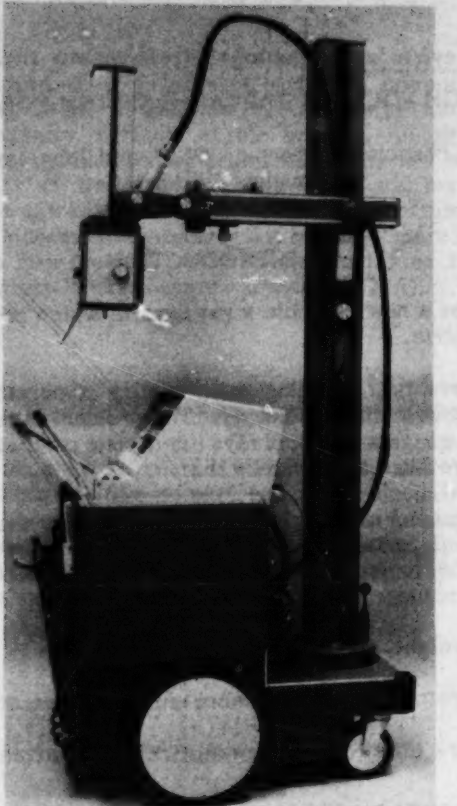
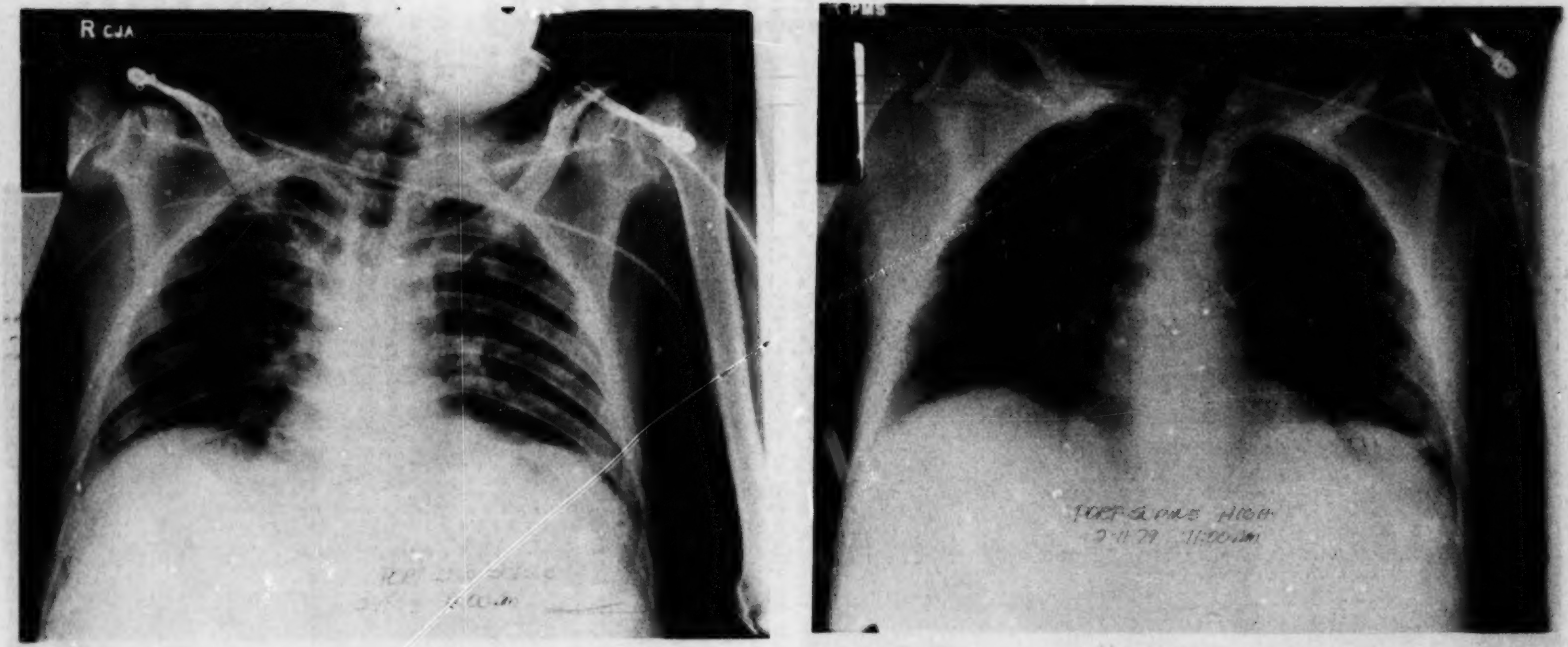


Figure 1. 240 kV phototimed chest portable.

Figures 2 and 3, obtained at Harbor General Hospital in Los Angeles, illustrate the differences between typical films obtained with conventional equipment (G.E. AMX 110) and with the new portable.

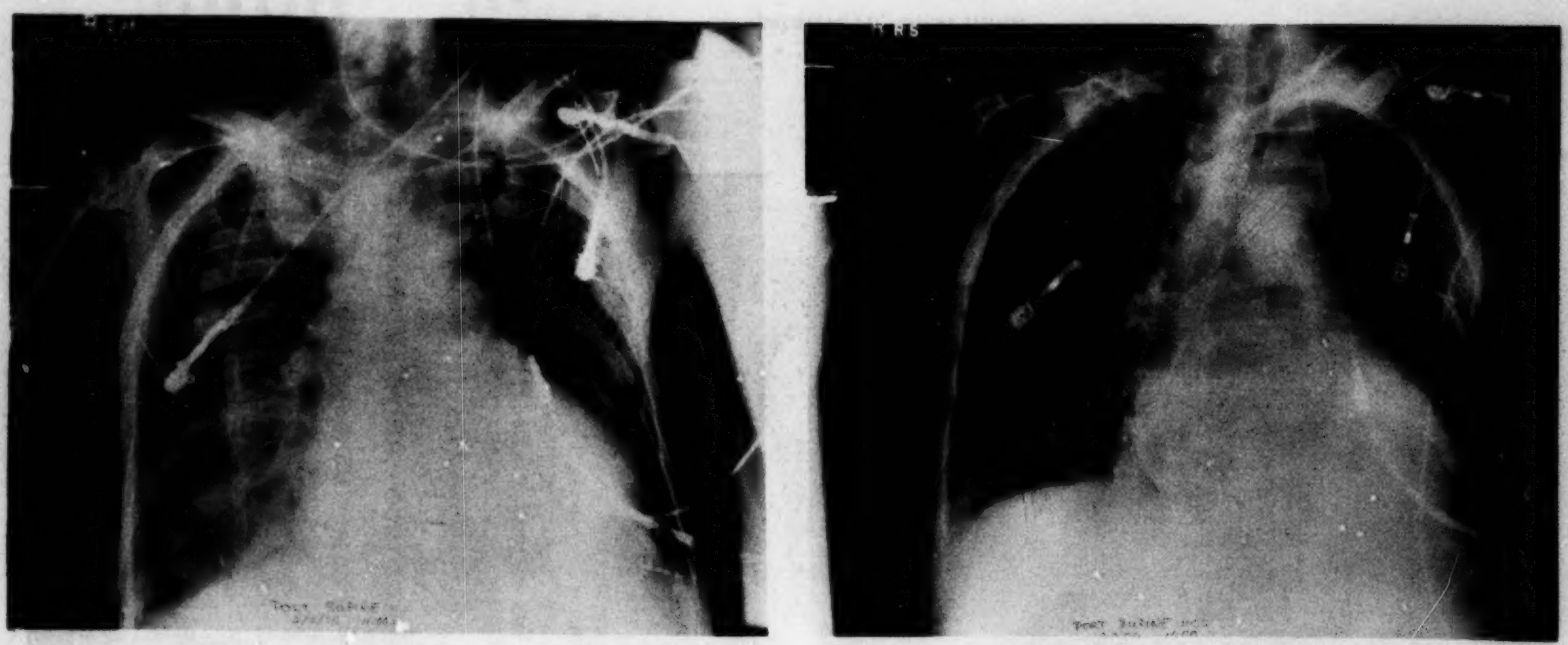
AUTOMATIC EXPOSURE CONTROL

A serious limitation to portable film quality is the difficulty of controlling exposure and the resulting frequency of film exposures poor enough that diagnostics are impaired and retakes necessary. Many ICU patients suffer from pathologies which greatly affect the radiolucency of the lungs, so that exposures determined from patient size alone can be in error by as much as ±50 percent, i.e., the optimum exposure for a given patient size may vary over a range of 3 to 1 depending on the size of the heart and the condition of the lungs. This problem is solved in the new unit by means of a compact remote automatic exposure control system attached to the film cassette. The total package is shown schematically in Figure 4.



R CJA
PMS
PORT SUPINE HIGH
2-11-79 11:00 AM

Figure 2. Comparative AP chest films.



(a) 90 kVp

(b) 240 kVp

Figure 3. Comparative AP chest films.

REMOTE AUTOMATIC EXPOSURE CONTROL

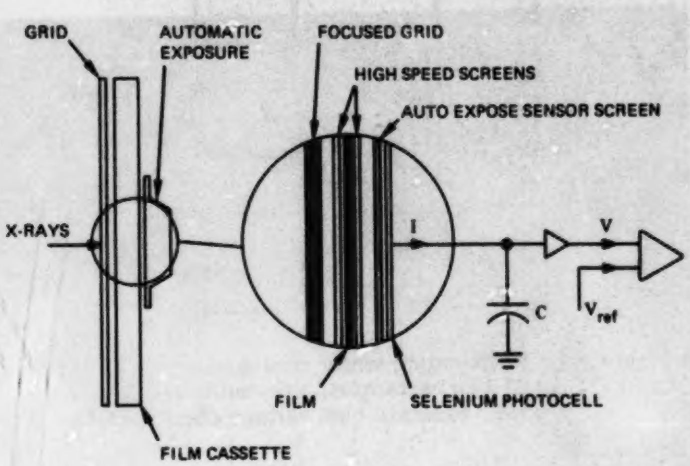


Figure 4. Film cassette holder with automatic exposure control schematic diagram.

The field emission x-ray source generates a train of submicrosecond pulses at the rate of 1,000 pulses per second. A PA chest exposure is typically on the order of 20 pulses. An appreciable fraction of the x-ray beam traverses the cassette and reaches the automatic exposure sensor screen. A selenium photocell converts the light from that screen into an electrical current, I, which is integrated and converted into a volgage, V. The x-ray intensities reaching the film and the automatic exposure sensor are in a fixed ratio, thus the voltage V is proportional to film exposure. After each pulse, i.e., every millisecond while the exposure progresses, the voltage V is compared to a reference voltage which is preset to represent the desired lung field density for the film/screen combination used. As soon as the output voltage reaches the reference voltage, the system is shut off before the onset of the next pulse. The accuracy of the system is within ± 5 percent, hence the film density is repeatable within a maximum ± 0.2 variation when allowance is made for variations in patient and positioning.

The typical phototimer sensor arrangement is shown by the dotted rectangles in Figure 5. The two smaller rectangles are normally used for PA chest films and their output is averaged. This arrangement presents two problems in ICUs. First, the 17-inch film dimension may be either vertical or horizontal depending on the size of the patient. Hence the sensor array should have fourfold symmetry so that its response is independent of cassette orientation. Second, a number of ICU patients have severe problems (e.g., lobectomy, hemorrhaging, severely enlarged heart) which may cause one lung to be partly or mostly opaque to x rays. With an averaging technique such problems result in the open lung field being badly overexposed as illustrated in Figures 6 and 7.

These two problems were solved in the new portable by using a symmetrical configuration of four independent square sensors. The integrated outputs of the four individual sensors are continuously compared and the largest of the four is retained and compared against the reference voltage. As long as at least one of the four sensors is in the open lung area, that sensor prevails and correct exposure is assured.

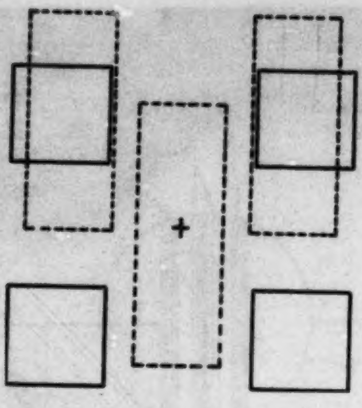


Figure 5. Phototiming sensor configuration
Dashed rectangles: conventional
Solid squares: high voltage chest portable

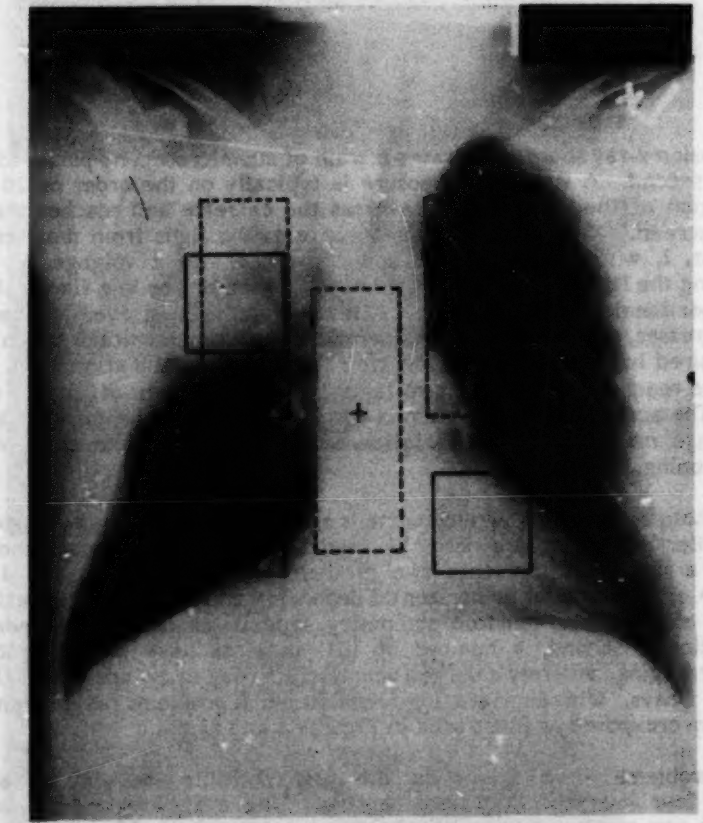


Figure 6. Sensor positioning on AP chest.

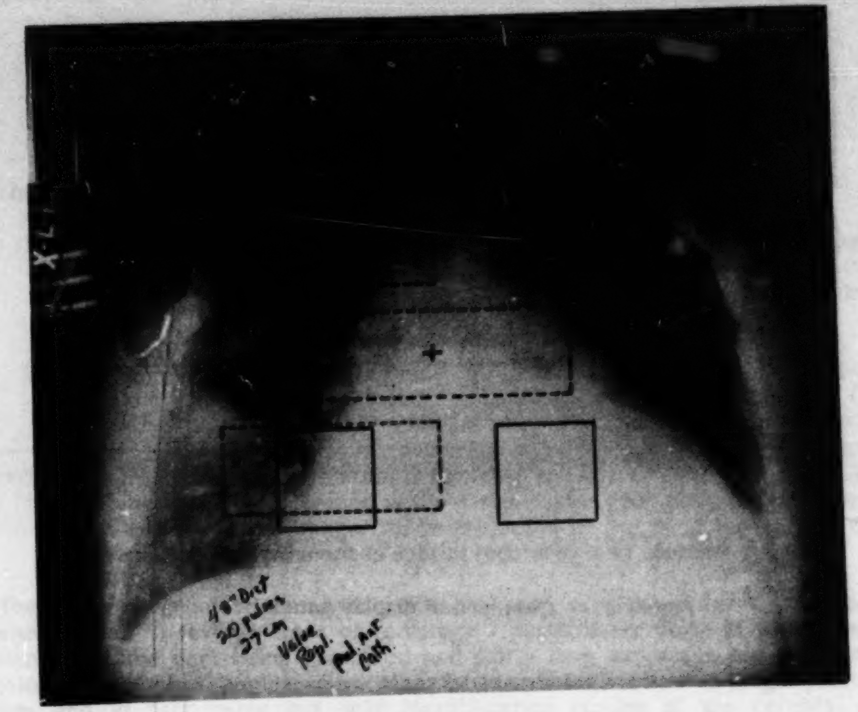


Figure 7. Sensor positioning on AP chest.

MICROPROCESSOR CONTROLS

For the first time in an x-ray portable, microprocessor based controls have been incorporated to simplify operation and reduce reliance on operator experience and training. As an alternative to the automatic exposure mode, the microprocessor can be used to calculate the correct exposure in a compute mode: upon keyboard entry of the patient size and the source-to-film distance, a technique chart stored in the system memory determines and displays the correct exposure. The keyboard display panel, shown in Figure 8, also prompts the operator through the correct operating procedure, displays information or error codes which verify correct operation or identify possible malfunctions, and can be used for more detailed functional equipment checks in a service mode.

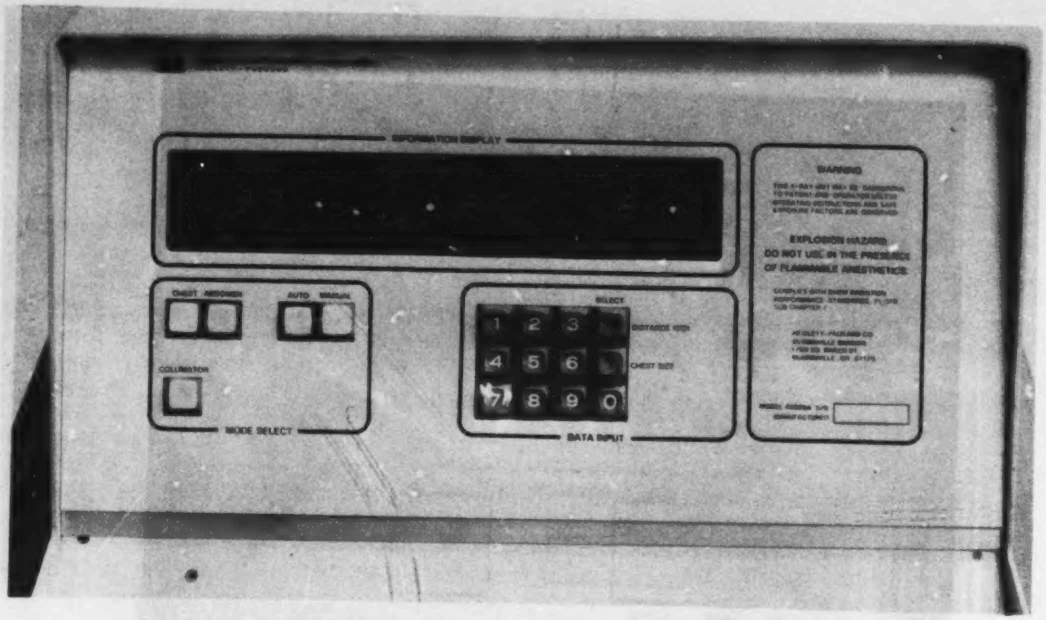

INFORMATION DISPLAY
CHEST
ABDOMEN
AUTO
MANUAL
COLLIMATOR
MODE SELECT
SELECT
1 2 3
DISTANCE (SID)
4 5 6
CHEST SIZE
7 8 9 0
DATA INPUT
WARNING
EXPLOSION HAZARD
DO NOT USE IN THE PRESENCE
OF FLAMMABLE ANESTHETICS
HEWLETT-PACKARD CO

Figure 8. Control and display panel.

MAGNIFICATION RADIOGRAPHY OF THE PEDIATRIC CHEST

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There are three principal ways of magnifying the radiologic image:

1. By radiographing the object in contact with a fine grain film and magnifying the image photographically. This technique can only be applied to very small parts - (e.g., fingers, wrists) and, because of the slowness of the film, requires a higher radiation dose.

2. The very slow fine grain film may be replaced by a much more sensitive recording medium such as the input face of an x-ray sensitive vidicon tube (1). This is currently an experimental technique which is suitable only for very small parts.

3. By projection magnification - utilizing a minute focal spot the object-to-film distance may be increased without significant geometric unsharpness developing and a much larger image can be projected on the same screen/film combination wih marked improvement in spatial resolution and contrast.

The value of geometric magnification has been established for many years and it is commonly used in several areas of radiology - particularly those in which there is little movement of the part being examined and fairly long exposures are possible, e.g. bone radiology and cerebral angiography. Many neuroradiologists now no longer perform contact studies and in fact carry out only magnification studies of the cerebral circulation. However, despite the fact that chest x-rays comprise between 40 and 50 percent of all roentgenology, magnification of the adult chest is rarely performed. The resultant image does yield more information, but there are some formidable logistic difficulties which deter the radiologist from using magnification as the first imaging modality. The principal one of these is that one would require huge specially constructed screen/film cassettes and special developing facilities. Magnification is therefore seldom used in adult chest studies except occasionally to obtain a better "second look" at some area which has already been identified as suspicious from the initial contact film.

The above objections do not apply to the tiny pediatric chest and the reasons for requiring magnification in these cases are quite compelling. First, the structures to be visualized are much smaller than in the adult chest and may be beyond the resolving capacities of the conventional 6-foot anode film distance chest imaging system. Second, the infant patient cannot hold his breath and motional blurring may occur unless a very fast screen/film combination is used - which in turn causes serious degradation of spatial resolution. Third, the very small infant chest is a low contrast structure and on a contact radiograph secondary scatter degrades the contrast still further. Fourth, the fast screen/film combinations now used require very little radiation to achieve adequate film blackening through the small lucent infant chest and quantum noise therefore becomes a serious problem. If geometric magnification is employed, many of these image degradations can be removed.

1. A large image (signal) is now thrown onto the same grainy (noisy) fast screen/film combination, increasing the signal-to-noise ratio (Fig. 1).

2. The air gap between the chest and the film clears up scattered radiation and markedly improves contrast.

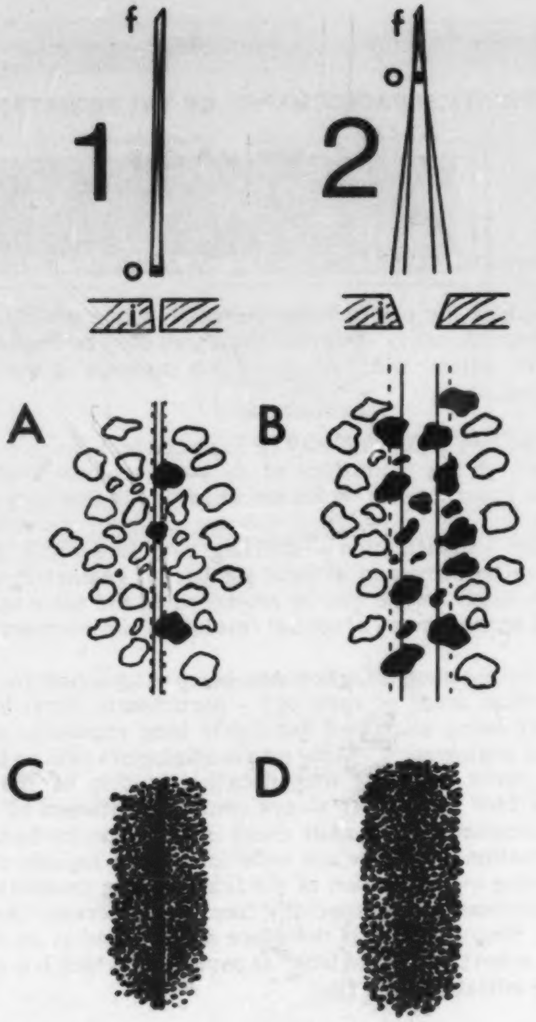

Figure 1

Column 1: **Contact Radiograph**

f = focal spot
o = object
i = image in air (of a wire)

The edge gradients of the image (i) in air are steep (well-defined) but when this small image (signal) falls on a very coarse-grained (noisy) film A, only one or two grains are energized and a poor image is produced. If i now falls on a fine grain film C, an excellent image is produced.

Column 2: Projection Magnification

The edge gradients of the image in air (i) are now less steep (poorer definition) but the image (signal) is much larger and the noise B is unchanged, i.e., an improved image (signal-to-noise ratio). A recognizable image is now produced. If i falls on D, however, the less well-defined image is faithfully recorded, i.e. a poorer image than that obtained by contact on a fine grain film.

(Illustration modified from ACR Monograph on Fluoroplanigraphy and Magnification Techniques: 1978)

3. Minute structures, previously difficult to see because they are beyond the range of resolution of the imaging system and because they are simply too small for the unaided human eye to detect, are now brought within the range of system resolution and the observer's vision.

5. The same amount of radiation is now incident on a smaller surface area, i.e., there is better quantum saturation per unit area and quantum noise therefore is less.

In order to keep motional blur at a minimum a rapid exposure still needs to be made which provides a lower limit to focal spot size (in order to achieve adequate heat loading). Currently electron-gun focused tubes are available with focal spots in the region of 100 µm; however, the heat-load capacity of such tubes is small and there is still a problem with achieving rapid enough exposures to freeze motion. These exposures should preferably be of not more than 1/100 of a second. With a 100 µm focal spot, very large magnification (6 to 8 times) would be possible with progressive increase in spatial resolution, but the increasing anode-to-film distance required makes it necessary to use exposure times which are too long to avoid motional blur. Until the heat-load capacity of micro-focal spot tubes can be increased, magnifications between 1.5 and 2.5 are usually employed with exposures in the range of 1/100 to 1/130 of a second. Clinically, the two times magnification (four times "area" magnification) makes a profound difference in the information content of the pediatric chest and permits diagnoses to be made which are impossible on standard contact films. The trade-off in obtaining this increased information is of course increased radiation dose (skin dose). For example, if the skin dose from a conventional radiograph covering an area of approximately 1,000 sq cm is 4.0 mR at 90 cm anode-object distance, a two times enlargement radiograph covering 160 sq cm would give a skin dose of approximately 25 mR, i.e., a sixfold increase. However, if one considers the total dose received, i.e., the integral surface dose (skin dose multiplied by field size), there is little difference in the total radiation dose (2). The gain in information content on the radiograph is, however, so dramatic and the skin dose so small (even on magnification) that the known advantages certainly outweigh the purely theoretical disadvantages of the increased skin dose.

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DISCUSSION

DR. CAMERON: For following patients in intensive care, it seems to me that phototiming works against you, because what you would like to do is reproduce exactly the physical conditions you had on the previous films. That is, you need very good quality control on the kVp, the exposure, and the processing. If you have that, and you see a change in lung density, you will be fairly confident it is real. With phototiming, you would have about the same optical density over the lungs and you might miss a change of lung density.

DR. ALTER: I might be pre-empting in part what Dr. Charbonnier is about to say, but there are many factors involved in getting reproducible registration of lung density. Some are physiological factors: depth of inspiration, patient position, the amount of soft tissue or lung edema; these may change from day to day. We have a book in our department where the technicians record the factors on every patient who has a portable examination so those factors can be used again the next day. The problem is that the films are still not the same. Phototiming is perhaps part of the answer, but I do not think it can be the complete answer.

DR. CHARBONNIER: I agree. However, I think that unplanned variations in exposure can be a great problem in making films comparable. A system which offers automatic exposure is probably the best way of getting consistency of density on serial films.

The Hewlett-Packard portable chest system can be used in any of three modes: automatic exposure control, computed exposure, or manual. The automatic exposure control mode will generally insure the optimum and consistent exposure desired for good diagnostic reading and recognition of changes. However, if desired, the actual exposure in milliseconds (which is displayed on the front panel) can then be used on subsequent films done in the manual mode to display gradual changes in the radiolucency of the lungs (e.g., due to a diffuse increase in lung fluid) that might be masked by systematic use of phototiming.

In the computed exposure mode, we can use technique charts stored in the memory of the microprocessor. We simply enter on the keyboard the distance from the film to source and the patient's size, as we judge it. The system will choose, display, and deliver the appropriate exposure time. All these options are available and might be used to good advantage. It is not a panacea, but it does broaden your options and may give you an opportunity to improve.

MR. FIELDS: I have a comment and a question. Dr. Alter mentioned some problems with catheters. The catheter manufacturers have specifications such that to be considered radiopaque the catheter must give a minimum 0.1 density unit variation at 80 kVp through 1.5 mm of aluminum. I agree that at much higher kVp you would most likely lose visibility.

Dr. Charbonnier, what is the duty cycle on your 240 kVp machine? How many films per hour or minute can we expect from it?

DR. CHARBONNIER: It can produce one film every two minutes, which is adequate for portable applications. For a short period of time, we can certainly exceed the duty cycle. I might say that the exposure depends on the film/screen combination but would typically be 15 to 25 milliseconds for an AP chest at 48 inches.

We did not really appreciate the problem of catheter visibility until we started getting field use in ICUs. I may summarize in this way--we do not seem to have any problem with endotracheal tubes, with nasogastric tubes or with the Swan-Ganz and CVP catheters. As a matter of fact, on the CVP catheter that goes through the heart, or the Swan-Ganz, we can really see the whole catheter through the mediastinum and through the heart, which is an advantage. The one type of catheter that has been giving us problems is the subclavian hyperalimentation catheter, which has several manufacturers. Some manufacturers use a small concentration of radiopaque material which is homogeneously distributed in the plastic tubing, the Bard or old Deseret catheters for instance, and this has been most difficult for us to image. Those catheters with a radiopaque stripe in them rather than the diffuse opaque material, such as the Sorenson or the new Deseret, present no imaging problem. When we go from 80 kVp to 240 kV, the barium or iodine which is the usual contrast medium loses contrast. At the same time, however, if the catheter is in the mediastinum, which is an area we are penetrating better, film response is improved and in some cases we can see the catheter as well as or even better than with a conventional voltage system. To summarize, with 2 out of 3 major subclavian catheters we do not seem to have a problem, and with other types of catheters we do not have a problem at all.

INFORMATION ENHANCEMENT AND DOSE REDUCTION IN CHEST RADIOGRAPHY: THE INTERGATED* RADIOGRAPHY SYSTEM

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Department of Radiology, Massachusetts General Hospital
and
Robert E. Wernikoff, Ph.D., Brattle Instrument Corporation

INTRODUCTION

Coronary artery disease may impair blood supply to part or all of the myocardium and eventually expresses itself in abnormal contractile pulsations of the deprived region of the heart. Incompetent heart valves and shunts between the left and right sides of the heart result in volume overloads and they also manifest themselves as pulsation abnormalities of the affected chambers and connecting blood vessels. Thus, a large class of diseases of the heart may be detected and assessed by imaging cardiovascular pulsations, and doing that is the main purpose of contrast ventriculography and of radio-labelled cardiac blood-pool studies. These methods, however, are not suitable for screening or for frequent follow-up. Conventional chest radiography is suitable, and it is indispensable for the evaluation of cardiac function, but it does not give direct information about cardiovascular pulsations. Showing how this information may be included in chest x rays, without perceptible interference with the ordinary purposes of chest radiography and without increasing radiation dose to the patient, is the purpose of this report.

The technique is called interlaced ECG-gated radiography or, more briefly, Intergated Radiography; the device that makes it possible may be attached to most existing chest x-ray units. Initial experience in using Intergated Radiography in place of the conventional admission chest x ray in more than 500 cases indicates that cardiovascular motion can be reliably assessed without apparent loss of quality of the image with respect to other thoracic structures. We have found excellent correlation between regional wall motion on Intergated films and ventriculograms in patients with coronary artery disease, and between aortograms and Intergated films in patients with aortic regurgitation (1,3-4).

MATERIALS AND METHODS

The new system, Intergated Radiography, is shown diagramatically in Figure 1. A movable mask is interposed between the x-ray source and the patient. The mask is an array of parallel strips of lead, each 3 mm wide and thick enough to block radiation in the diagnostic energy range, alternated with equally wide spaces. The effect of making an exposure with the mask in place is that alternate 3 mm horizontal strips of film are exposed while the shielded strips remain unexposed. After the first exposure, the mask is advanced 3 mm to its second position, in which its lead strips shield the previously exposed film. The second exposure then fills in the previously unexposed areas of film. The film therefore shows an interlacing of the two component exposures, in which stationary anatomical structures appear substantially as they would on an ordinary film, whereas moving structures, such as the borders of the heart and great vessels, have a serrated appearance that easily permits measurement of the amplitude of pulsation.

The mask and the x-ray machine are under control of a physiological synchronizer. By means of electrodes applied to the patient, the synchronizer detects the patient's electrocardiogram (ECG), from which it determines the timing for triggering signals that turn on the x-ray tube. Thus,the two gated exposures are made at times that approximate end-systole and end-diastole. The first exposure is triggered at end-diastole by the R wave of the QRS complex and the second at end-systole after a delay that is a linear function of the previous R-R interval. The ECG is recorded with a strip-chart recorder whose event-maker indicates the exposure intervals. In the installation reported here, the synchronizer is interfaced with an x-ray machine operated at 130 kVp and 400 ma and a focus/film distance of 72 in. With these factors, exposure intervals range from 40 to 80 ms and average about 50 ms. In this range of exposure intervals, the ventricular volumes at end-systole and end-diastole are relatively stationary during exposure and the cardiac borders are sharply defined. Each study is completed in one cardiac cycle.

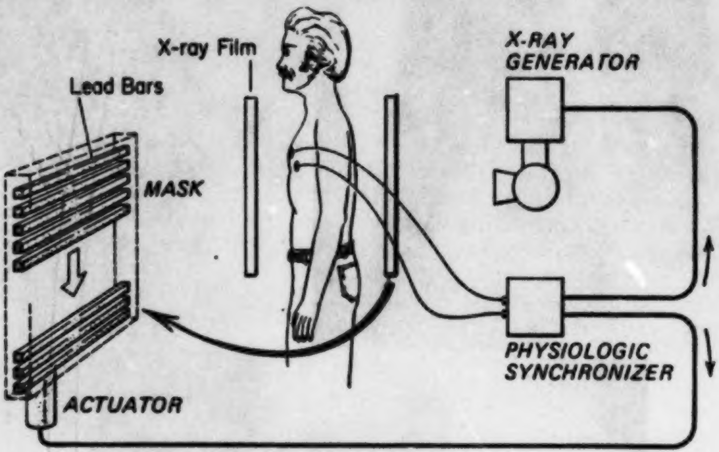

Figure 1. Diagram of the Intergated Radiography System interfaced with a conventional chest radiography unit. The interlacing mask was originally positioned between patient and film, and most of the illustrations in this paper were done in that manner. More recently, we have found that the mask may be placed between source and patient, as shown here, without perceptible change in image quality and with reduction of patient dose down to or below the level of conventional radiography.

Intergated Radiography requires two exposures per film but, since the mask itself shields one-half the image area of the patient during each exposure, the absorbed dose can be made equal to that of conventional radiography. In fact, if the structure that supports the mask between source and patient is equipped with additional shielding all around the mask, patient dose may be reduced to below that of conventional chest radiography. The Intergated Radiography System is simple and relatively inexpensive, and is designed to be interfaced with standard radiography units.

Figure 2A shows a PA view done with this system. Note the good detail of thoracic as well as of cardiovascular structures. (Image quality may be lost in printing. Photographic copies may be obtained, while the supply lasts, from Dr. Wernikoff at 243 Vassar Street, Cambridge, Massachusetts 02139.) As shown in Figure 2B, lung detail is essentially unaffected by the interlaced gating process. The detail view of a portion of the heart border in Figure 2C shows by its strips how the end-diastolic and end-systolic images are interleaved, or interlaced, through the action of the mask. If, as shown in Figure 2D, corresponding strips are imagined to be connected, the resulting curves represent the heart border at end-diastole and end-systole, and distances measured perpendicularly to the curves represent cardiac contractions and can be accurately measured by caliper. Marginal markings at the edge of the film (not shown in the figure) identify the strips that form the systolic image. Another normal study is shown in Figure 3.

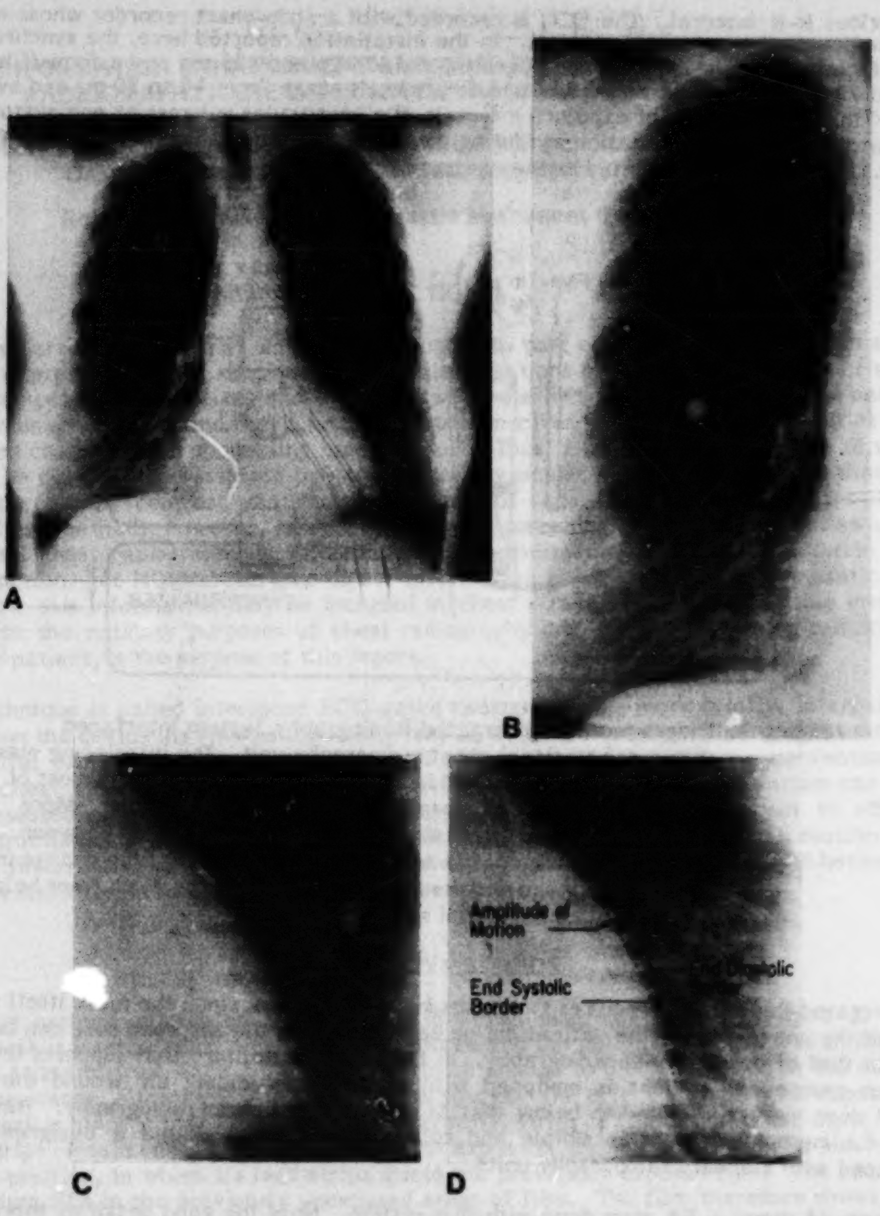
A
B
C
D
Amplitude of Motion
End Diastolic Border
End Systolic Border

Figure 2. (A) Normal postero-anterior ECG-gated, interlaced roentgenogram, showing good detail of the heart and other thoracic structures. (B) Detail view of the right lung, showing that image quality is essentially unaffected by the interlaced gating process. (C) Detail view of a portion of the heart border, approximately 1:1, showing the alternation of end-diastolic and end-systolic image strips. (D) Same as (C) with heart borders at end-diastole and end-systole drawn in, and illustrating that amplitudes of cardiac contractions should be measured perpendicularly to the cardiac borders.

A

B

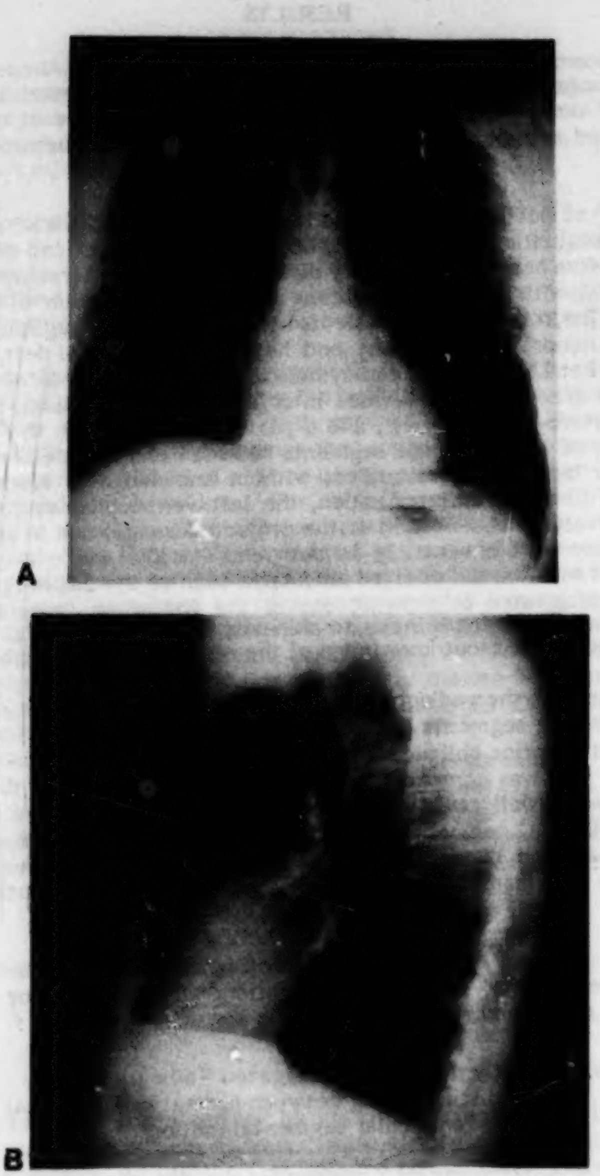

Figure 3. (A) PA view and (B) lateral view, both showing normal motion.

In the normal heart, amplitude of motion of the left ventricle is small, with a mean of 3 mm, but easily measured. The normal aorta moves approximately 2 mm. The motion of individual segments of the left ventricle can also be studied in lateral and oblique views. Of course, the interventricular septum and, in most cases, the inferior wall are not shown with this technique.

RESULTS

To date the system has been used on more than 500 cases. Almost all were cardiac admissions and a majority subsequently underwent cardiac catheterization; thus, system evaluation is based on objective data (1,3-4). Consideration here of one early series of 35 patients suspected of coronary artery disease may serve to illustrate methodology and typical results (1).

All 35 patients had postero-anterior and lateral Intergated chest x rays upon admission to the hospital for evaluation of chest pain, and all were catheterized on the day following admission. Twenty-five had coronary artery disease as shown by coronary arteriography and left ventriculography, with at least 50 percent reduction in diameter of one or more major coronary arteries. The remaining ten patients had coronary arteriography which was either normal or showed minor abnormalities, and had entirely normal left ventriculography. Motion of the left ventricle between end-systole and end-diastole was evaluated as follows. The left ventricular free wall was divided into four segments: basal, lateral, and apical segments on the postero-anterior view, and a posterior segment on the lateral view. The amplitude of motion of each of these segments between end-systole and end-diastole was measured by caliper by one of the authors without knowledge of subsequent ventriculographic findings. Following catheterization, the left ventriculograms were analyzed for wall motion subjectively by observation of the projected image, and in addition objectively by tracing the outline of the ventricle in both end-diastole and end-systole, and superimposing the tracings with the aid of fixed landmarks such as the diaphragm and a rib. Each segment was then designated as showing one of the following: normal motion; mild to moderate hypokinesis; severe hypokinesis to akinesis; or dyskinesis. This analysis was done by another of the authors, without knowledge of the radiographic findings.

The three segments of the radiographic postero-anterior view were compared with the corresponding neighboring segments on the right anterior oblique ventriculogram. In those cases in which a left anterior oblique ventriculogram was available, the posterior segment on the lateral chest film was compared with the posterior segment on the ventriculogram. The septum and inferior wall could not be compared since these were not visible on the radiographs. A total of 112 segments were compared, 80 segments in the 25 patients with coronary artery disease and 32 segments in the 10 control patients without coronary artery disease. Data from the latter served to define normal ranges of motion for Intergated Radiographs.

The data were used to estimate the effectiveness of Intergated Radiography in distinguishing between normal and abnormal motion. The results for the 80 segments studied in this series are given in Table 1.

Table 1. Effectiveness of Intergated Radiography in distinguishing between normal and abnormal left-ventricular segmental motion

	negative by roentgenography	positive by roentgenography	totals
normal by ventriculography	29	2	31
some degree of asynergy by ventriculography	2	47	49
totals	31	49	80

$$\text{Sensitivity} = \frac{\text{detected positives}}{\text{all positives}} = \frac{29}{31} = 94\%$$

$$\text{Specificity} = \frac{\text{detected negatives}}{\text{all negatives}} = \frac{47}{49} = 96\%$$

APPLICATIONS

Intergated Radiography has been found very useful in initial examinations of patients with possible heart disease as well as for serial follow-up of patients with known cardiac conditions. It has been particularly useful in coronary artery disease, valvular heart disease and congenital heart disease.

An example of a patient with coronary artery disease is shown in Figure 4A. There is normal contraction in the basal area of the left ventricle, diminishing in the lateral segment, and there is dyskinesis at the apex. The fact that the apex moves outward in systole can be verified on the original film by referring to the marginal markings, which identify the systolic strips. The diagnosis was confirmed at ventriculography.

Intergated Radiography is also very useful with the group of cardiac conditions characterized by volume overload affecting the heart or great vessels, such as aortic regurgitation. This group includes patients with regurgitant valve lesions and some of the congenital left-to-right shunts. The findings that suggest volume overload are (1) enlargement, and (2) increased amplitude of motion of the wall of a cardiac chamber or a vessel, reflecting an increased stroke volume. For example, a patient with aortic regurgitation might be expected to show a volume overload pattern affecting the left ventricle and the aorta and that is just what Figure 4B shows for a patient with severe aortic regurgitation. The aorta shows 7 mm motion, compared with the normal 1-2 mm, and the enlarged left ventricle shows up to 7 mm, compared with the maximum of 4 mm we have found for normal-sized ventricles. In a study of 17 patients with moderate to severe aortic regurgitation, all confirmed by cardiac catheterization and aortography, we have found a highly significant difference in amplitude of aortic and left ventricular motion between patients with moderate to severe aortic regurgitation, and the normal controls. Our findings are summarized in Table 2.

Table 2. Intergated Radiography measurements in aortic regurgitation (in mm) (3)

Aortic Regurgitation	Number of Patients	Maximum Left Ventricular Motion		Maximum Aortic Motion	
		Range	Mean	Range	Mean
None	10	2-4.5	3.3 ± 0.8	2-3	2.1 ± 0.3
Moderate	6	4-8	6.2 ± 1.6*	3-6	4.3 ± 0.9*
Severe	11	5-9	6.3 ± 1.3*	3-8	5.4 ± 1.5*

*$p < 0.001$

Mitral regurgitation is characterized by volume overload of the left ventricle and left atrium. Of course, the amplitude of wall motion also depends on such factors as the condition of the myocardium, its systolic function and compliance. Figure 4C shows a typical example of the changes that occur with mitral regurgitation. The left ventricle and left atrium are enlarged, and both chambers show an increased stroke volume.

Another patient (Fig. 5A) had severe paravalvar aortic regurgitation, demonstrated on her LAO preoperative Intergated Roentgenogram by the abnormal tilting motion of the prosthetic valve ring, measured on this film at 11° (compared to a measurement of 10° on a cine examination). This is an abnormal measurement. Angular motion of an aortic prosthetic valve greater than 6° suggests separation of the prosthesis from its bed (5). Following surgery (Fig. 5B) aortic motion returned to normal (2 mm). However, left ventricular motion remained increased in amplitude (7 mm), and the chamber showed further increase in size. In addition, the left atrium became enlarged and showed exaggerated stroke volume as well (3 mm, in the lateral view). The patient had developed an inferior myocardial infarct intraoperatively, with papillary muscle dysfunction and mitral regurgitation, all confirmed at catheterization.

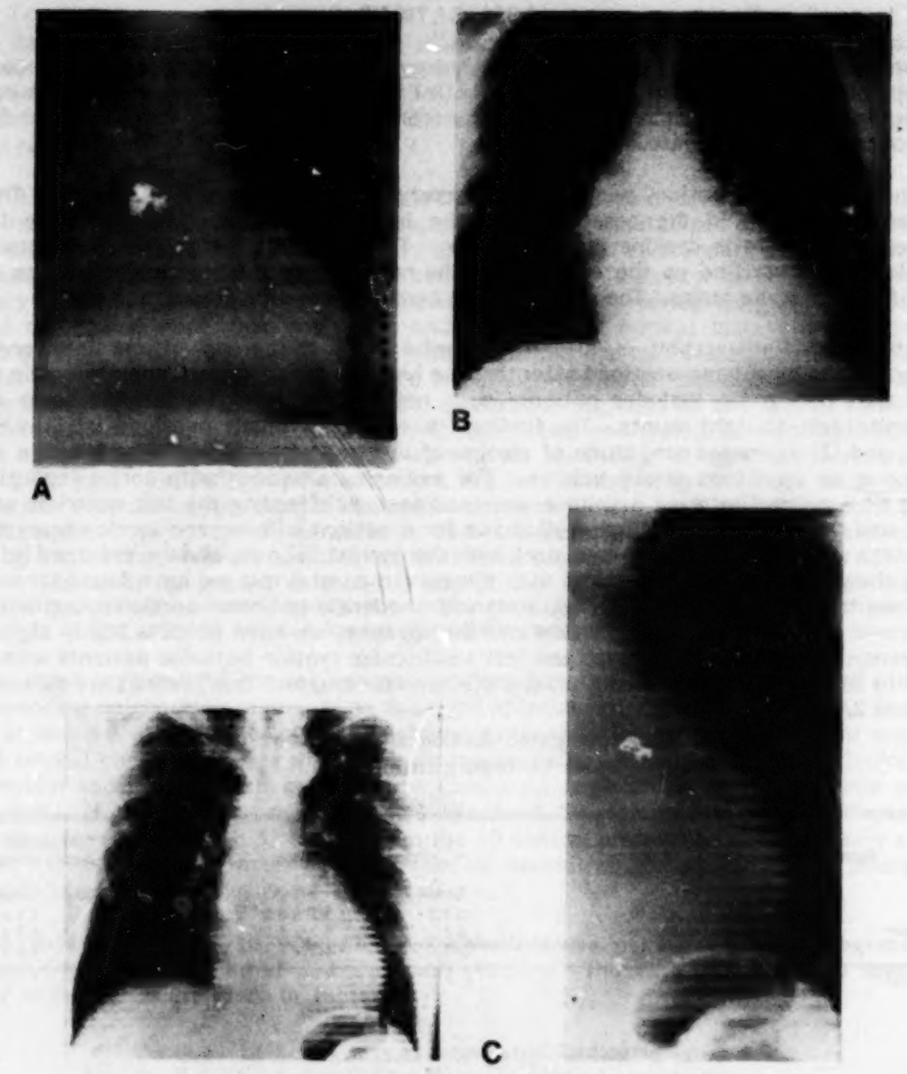

Figure 4. (A) Coronary artery disease. Normal contraction in the basal area of the left ventricle, diminishing in the lateral segment; dyskinesis at the apex. The black dots mark the systolic strips. (B) Severe aortic regurgitation. (C) Mitral regurgitation. Left ventricle and left atrium are enlarged, and both chambers show an increased stroke volume.

The results of examinations of patients with congenital left-to-right shunts are particularly interesting. These patients consistently show an increase in amplitude of motion of the pulmonary artery; in Figure 5C the measured amplitude is 4 mm. Figure 5D shows similar findings in another patient. The lateral view shows increased amplitude of motion of the large right ventricle, indicating that that chamber is volume overloaded. Both patients have inter-atrial septal defects.

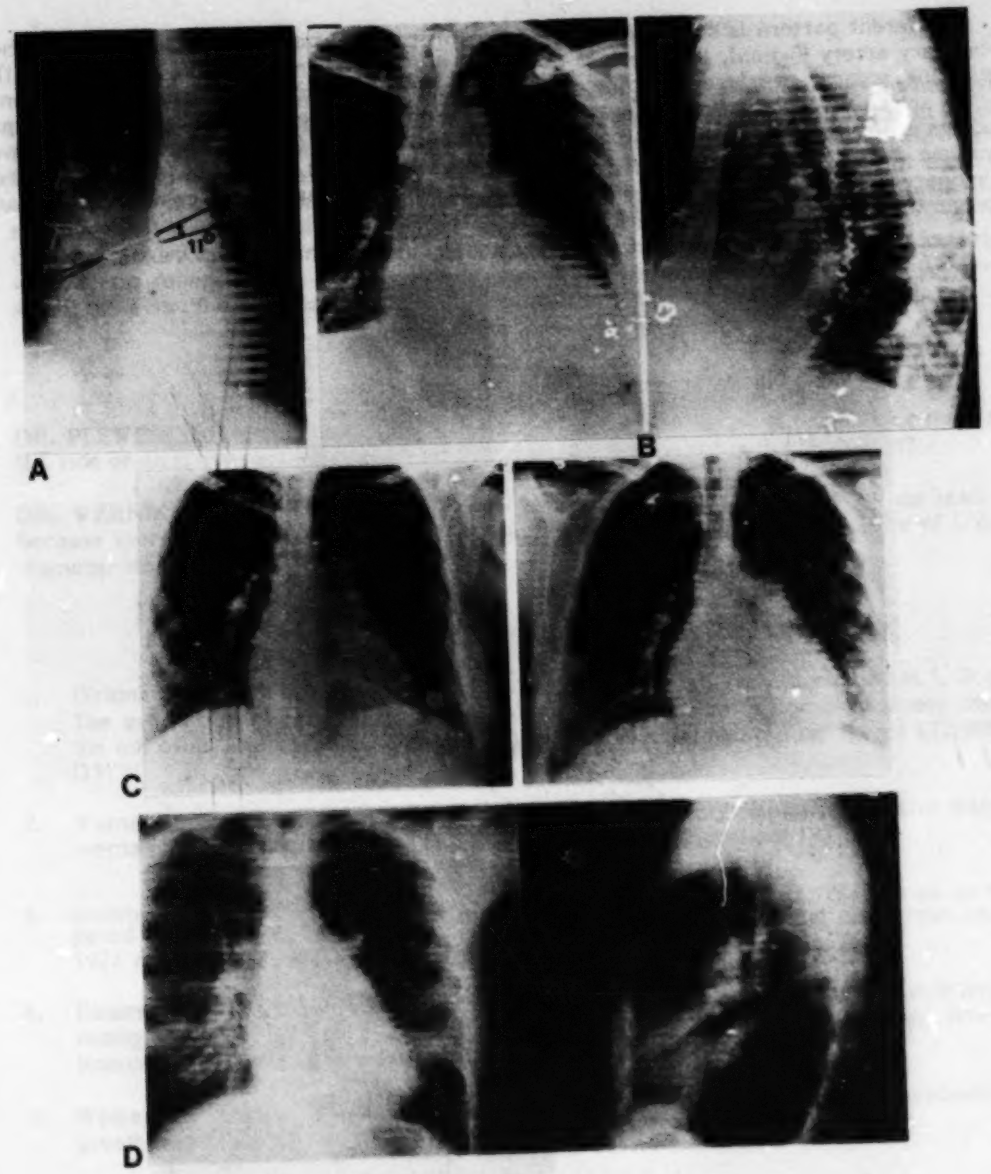

Figure 5. (A) Aortic regurgitation. LAO view showing abnormal motion of the prosthetic valve ring suggestive of separation of the prosthesis from its bed. (B) Same patient as in 5A, after surgery. Aortic motion has returned to normal. Increased size and motion of left ventricle, and enlarged left atrium with exaggerated stroke volume now suggest mitral regurgitation. (C) Atrial septal defect. Conventional film on the left, Intergated film on the right, showing increased motion of the pulmonary artery. (D) Atrial septal defect. Lateral view shows increased motion of right ventricle, indicating volume overload.

A different pattern is shown in Figure 6A. This patient shows increased motion of the pulmonary artery (3 mm), the aorta (3 mm), and the left ventricle (5 mm, posterior wall), indicating volume overload affecting those structures, a combination suggesting patent ductus arteriosus. This was shown at catheterization to be the correct diagnosis. Similar findings could also be caused by a combination of lesions (Fig. 6B). This patient also shows a volume overload pattern affecting the left ventricle (6 mm), the aorta (6 mm), and the pulmonary artery (3 mm), but in this case the findings are due to the combination of an interventricular septal defect, and aortic regurgitation.

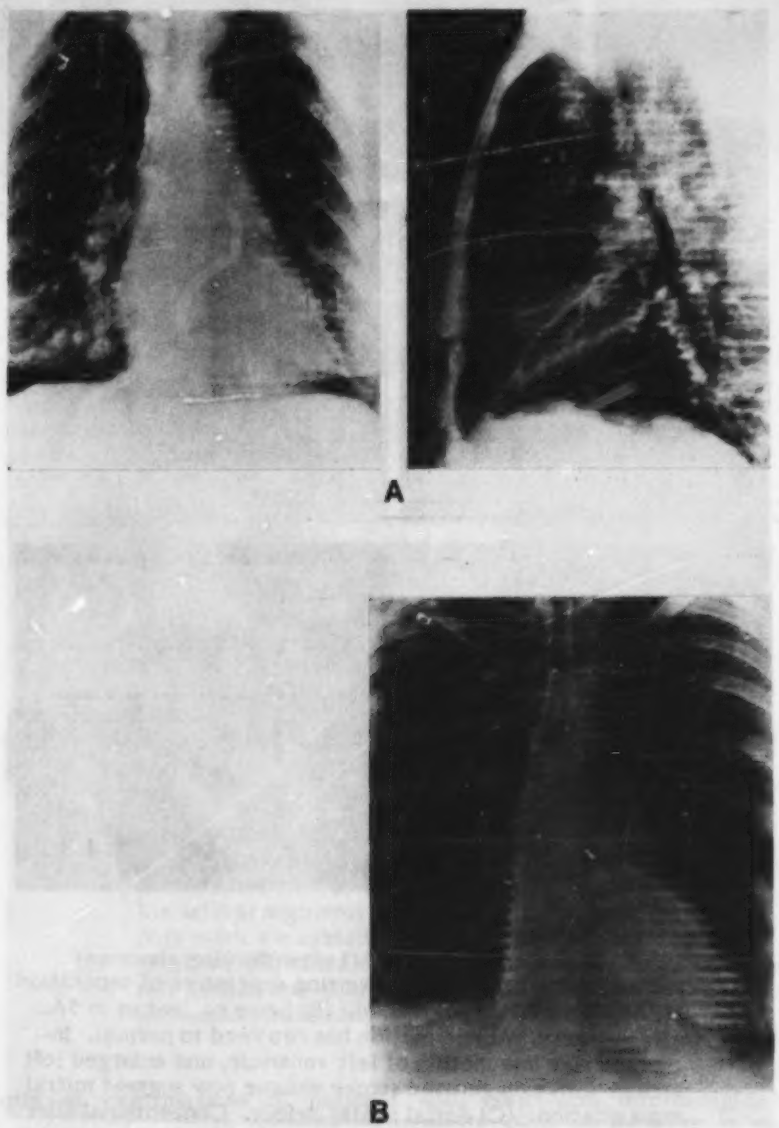

Figure 6. (A) Increased motion of pulmonary artery, aorta and left ventricle suggestive of patent ductus arteriosus. (B) Volume overload pattern affecting left ventricle, aorta and pulmonary artery due in this case to the combination of interventricular septal defect and aortic regurgitation.

SUMMARY

We have described a simple, inexpensive technique for increasing the information gained from the plain chest roentgenogram. Initial experience in using this technique in place of the conventional admission chest x ray in more than 500 cases has indicated that cardiovascular motion can be reliably assessed without apparent loss of the quality of the image with respect to other thoracic structures. We have found excellent correlation between regional wall motion on the Intergated chest x ray and ventriculograms in patients with coronary artery disease, and between aortograms and the Intergated chest x ray in patients with aortic regurgitation. Our series of patients with congenital left-to-right shunts is small; however, the technique appears very promising for their evaluation as well.

QUESTION

DR. PLEWES: Concerning those lead bars that slip back and forth--if you look at them from the side or on cross-section, are their boundaries sharp or are they milled round?

DR. WERNIKOFF: Actually, it turned out that it was most practical to use lead wire because there happened to be lead wire of the right size easily available. Wire of 1/8-inch diameter was used in Lucite® slots.

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REAL TIME UTILIZATION OF QUALITY ASSURANCE DATA TO OPTIMIZE CHEST RADIOGRAPHY; A PATIENT POSITIONING SYSTEM FOR CHEST RADIOGRAPHY—A PRELIMINARY REPORT

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University of Washington

We have heard many excellent talks about the need for optimizing chest radiography. We have learned that the image must be appropriate for the task specified, that the radiograph must be properly exposed for either the lungs or the mediastinum, and that visualization of lesions and other pathology is compromised by overlying and underlying structures.

Today, I want to discuss two issues which bear on these matters: first, I will explain briefly how to insure reproducible films in spite of small variations in processing conditions, and second, I will present a preliminary evaluation of a patient positioning system. Both issues will be approached using perturbation methodology. In this method, it is assumed that an optimal radiograph has somehow been defined, and the effect of changing one factor at a time is determined. In this way it is possible to order the factors in terms of the magnitudes of their effect on a chest radiograph. It will be shown that controlling only a few of the many possible factors will lead to highly reproducible chest images.

Ideally, all components of a radiographic system should be highly consistent and reproducible. However, in older institutions, surgical suites, small clinics, or private offices, conditions may be such as to make this an impossible goal. Proper corrective action may be exceedingly expensive.

Nevertheless, high quality, consistent radiographs can still be obtained if the instantaneous physical conditions are known. These conditions are known if a good quality assurance program has been established. While many parameters can be considered in the definition of a high quality radiograph, the two that I will discuss here are optical density and radiographic contrast. Other parameters can be controlled in a similar manner.

Optical density and radiographic contrast are dependent on many parameters. The most succinct presentation of the film-related parameters is an H & D curve. It is well known that many parameters affect the H & D curve. One parameter is temperature. Its effect is shown in Figure 1. As can be seen, the dominant effects are changes in contrast and speed, plus an increase in base and fog and maximum density. These changes are well known, but only the effect on speed has previously received much attention.

Optical photographers have long been aware of the changes in speed, contrast, and fog levels. They work hard for consistency; while they can compensate for speed changes, they cannot compensate for contrast changes.

We are more fortunate. We can do all that optical photographers can do, but in addition we can compensate for contrast changes, because we can control subject contrast through proper choice of kVp and filtration, as has been discussed at this symposium.

We have considered the variables shown in Figure 2. Optical density and radiographic contrast were taken to be functions of these variables. Once these variables were determined, total derivatives could be obtained (Fig. 3). The problem becomes one of estimating the necessary partial derivatives. This can be done either empirically or theoretically. If all of the processing differentials are known, then we are left with two equations in two unknowns.

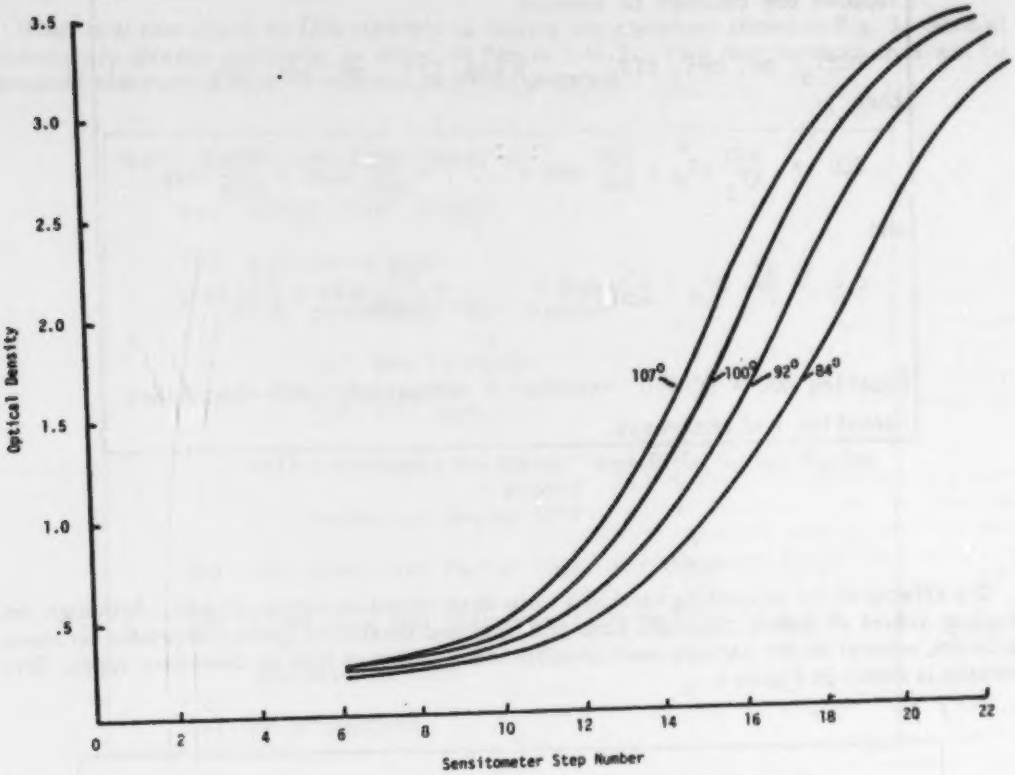

Figure 1

VARIABLES

- Processing
  - Film manufacturer (13)
  - Chemistry manufacturer (3)
  - Developer Temperature (4)
  - Wash Temperature (4)
  - Developer pH (3)
- Radiographic
  - Screen type (10)
  - mAs
  - kVp

Measured parameters from H and D Curve:

Base and Fog $\overline{G}$ Speed D_{max}

Figure 2

Suppose one chooses to express

$$OD(T_d, pH, mAs, kVp, \ldots) \text{ and } C(T_d, pH, mAs, kVp, \ldots),$$

then

$$dOD = \frac{\partial OD}{\partial T_d} dT_d + \frac{\partial OD}{\partial pH} dpH + \ldots + \frac{\partial OD}{\partial mAs} dmAs + \frac{\partial OD}{\partial kVp} dkVp$$

and

$$d\,C = \frac{\partial C}{\partial T_d} dT_d + \frac{\partial C}{\partial pH} dpH + \ldots + \frac{\partial C}{\partial\, mAs} dmAs + \frac{\partial C}{\partial kVp} dkVp$$

Equating $dOD = dC = 0$ results in radiographs with consistent densities and contrasts.

Figure 3

The effects of the processing variables were determined sensitometrically. Although the absolute values of speed, contrast, base plus fog, and maximum density depended on many variables, several of the partials were sensibly independent of film or chemistry types. One example is shown in Figure 4.

Masterson, et al have shown that the change in optical density caused by changes in developer temperature is relatively independent of the film and developer used. They found that

$$\frac{\Delta\ \text{Speed}}{\Delta\ \text{Temperature}} = 0.048 \pm 0.004 \quad {}^{\circ}F^{-1}$$

We have repeated their work, and expanded it to include 13 types of film and 3 types of developers. We found that near 92°F,

$$\frac{\Delta\ \text{Speed}}{\Delta\ \text{Temperature}} = 0.050 \pm .006 \quad {}^{\circ}F^{-1}$$

and that

$$\frac{\Delta\ \text{Optical Density}}{\Delta\ \text{Temperature}} = 0.02 \quad {}^{\circ}F^{-1}$$

Figure 4

What now remains is to find methods of solving the equations shown in Fig. 3. Several methods are already available, as shown in Figure 5 (1,2). Two new methods that will be discussed today are a slide rule (2) and an HP-67 program.

Means for Processor compensation:

- (a) DuPont "BIT" system
- (b) Hand calculator
    - (i) programmed "BIT" system
        - (a) mAs variable
        - (b) kVp variable
    - (ii) programmed Arrhenius' equation, using Taylor expansion around 92°F
- (c) mAs correction factor from sensitometric tests
- (d) Slide rule for adjusting film density to match any desired film
- (e) HP-67 program

Figure 5

The slide rule is constructed under the assumption that only the speed changes with temperature. From Figure 1, or Figure 4, one finds $\partial \log_{10} E/\partial T$ or $\partial mAs/\partial T = 0.06$ mAs (mAs/°F). Using this as a vernier on logarithmic axis, and sliding them together over the graph paper, one finds it a trivial matter to determine a new technique which will produce the same optical density as the standard technique at 92 °F. An example is shown in Figure 6.

Alternately, one can fit the empirically determined partials into an HP-67 calculator, along with "standard" values for such parameters as temperature and pH. If all but two of the differentials are known, e.g., dT_d and dpH, and only dkVp and dmAs are unknown, then the equations in Figure 3 degenerate into two equations in two unknowns, and can be solved for the values of dkVp and dmAs which make $dC = dOD = 0$.

The differentials are determined by having the technologist enter the instantaneous conditions into the calculator; the new kVp and mAs are then calculated. The technologist uses these modified settings for the next exposure. The results of this method are shown in Figure 7.

In addition to employing proper radiographic technique, we have learned at this symposium that patient positioning is vitally important. PA, lateral, and oblique views should be taken with known source-patient-receptor geometry. While this will not overcome problems of conspicuity, it will insure that the complexities of the image will be made stationary from one film to the next. Even the suggested use of stereo to shift structures to improve visibility requires precise a priori knowledge of the source-patient-receptor orientation.

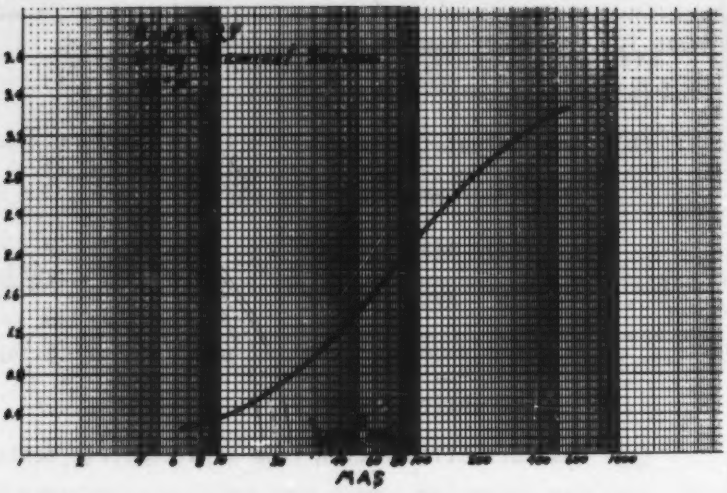

Figure 6

Temperature	mAs From Tech. Chart	Resulting OD	Corrected mAs	Resulting OD
84°F	40	.92	64	1.20
89°F	40	1.12	50	1.20
92°F	40	1.26	40	1.26
96°F	40	1.62	25	1.18
100°F	40	1.78	22	1.25

Figure 7

Assuming that the projections of all body structures must be constant to ± 2 mm, simple geometry indicates that the following must be obtained. (A distance of 72 in and a 12:1 focused grid are assumed. Note that all of these must simultaneously be attained. Failure to reach any one of these criteria results in more than ± 2 mm variation in projection.)

- **a. The arms must be extended overhead, with the hands reproducibly located to ± 3 mm.**
- **b. The shoulders and chest must be rotated reproducibly to ± 2°.**

c. The x-ray tube height must be reproducible to ± 1 cm.

d. The x-ray beam must be reproducibly angulated to ± 2°.

e. The center of the x-ray beam must be in the same place on the patient to ± 1 cm.

f. Lung inflation must be constant to ±6 percent in volume.

g. To correct for diurnal changes in stature, successive exposures must be made within two hours of the same time of day on successive days.

Several methods of achieving these tolerance limits have been proposed. They include mechanical systems sold by x-ray vendors and laser light systems described in the previous talk. These methods have met with various degrees of success.

The most difficult parameter to control is lung inflation. Several methods that might be considered employ strain gauges, linear transducers, spirometers, thermistors taped to the upper lip, magnetic and electrical measurements of conductivity, ranging devices, and ultrasound. None of these methods appears to be practical, and none are as simple as having the technologist demand that the patient take the deepest possible breath.

Positioning of the hands to ± 3 mm can be accomplished through careful use of standard extension devices available through most x-ray manufacturers. Patient rotation can be controlled by use of a panel low in absorption of both light and x rays, such as Lexan® (Fig. 8), to provide a support for the patient and to ensure an erect posture. Positioning and angulation of the x-ray tube can be done using tape marks or detents. The x-ray beam can be repositioned using either a spinal landmark or by marking the patient with an indelible marker at the center of the light field.

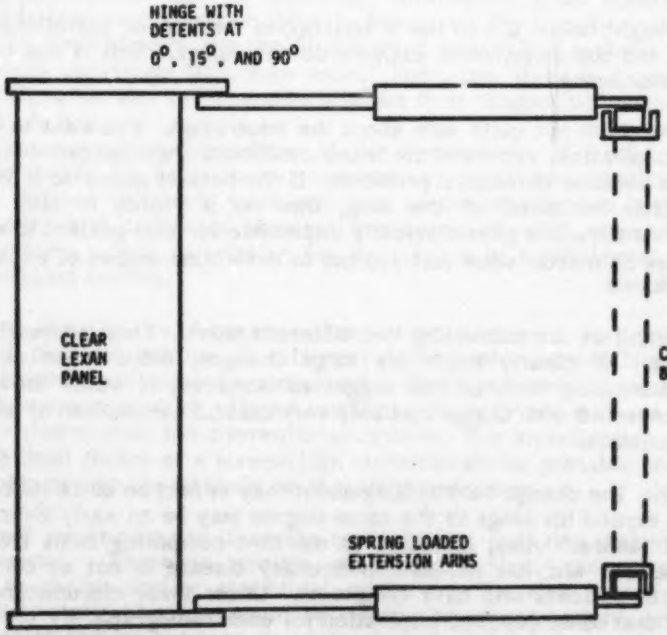

Figure 8

However, due to the problems of measuring inflation, and due to the problems of orienting patients who have excess amounts of fat, it does not appear possible to meet all of the positioning tolerance limits at all times. It may be that the present methods are close to an optimum trade-off between accuracy and practicality.

ACKNOWLEDGMENT

This work was supported in part by NCI Contract NO1CN45158.

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DISCUSSION

DR. NELSON: Are you marking your patient's skin at all?

DR. MOORE: No.

NELSON: That might help. If you use a felt-tipped marker, or something else relatively sanitary and not too obtrusive, most patients do not object. This is one method we have used for repeat positioning.

DR. GREENSPAN: I am not quite sure about the inspiration. You want to get films at the same degree of inspiration, yet there are many conditions that can prevent this. Films are taken on patients because of medical problems. If the patient goes into a little bit of heart failure, has a little "wetness" of the lung, then he is unable to take the same deep inspiration as previously. It is physiologically impossible for that patient to expand his lungs to the same degree no matter what test you use to determine degree of expansion. His lung compliance is reduced.

DR. MOORE: I think we are addressing two different points. I was addressing visualization of small changes. If clearly there are large changes, there is no need to compare radiographs; if something such as you suggested occurred, it would show up as a large change. I was concerned with things that only very careful comparison of one radiograph to another would elucidate.

DR. GREENSPAN: The change I am talking about may in fact be quite subtle; the fact that a patient cannot expand his lungs to the same degree may be an early indication that he is going into heart failure. Also, it seems to me that comparing films from one week to another for a patient who has no cardiopulmonary disease is not so commonly done as comparing films of patients who have symptoms. Under these circumstances, you cannot always duplicate that same depth of inspiration for each radiograph.

USE OF ANATOMICAL CRITERIA IN SCREEN/FILM SELECTION FOR PORTABLE CHEST X-RAY PROCEDURES

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The selection of a screen/film system for portable chest x-ray procedures is often a difficult choice. In particular, the advent of fast screen/film systems, many of them using rare-earth phosphors, has compounded the task facing the physician who must make a decision based on the diagnostic trade-offs between motion unsharpness, spatial resolution (including geometric unsharpness), contrast, noise (graininess or quantum mottle), and patient exposure. Whereas the faster systems may offer advantages for reduced motion unsharpness (particularly for the limited radiation output of certain portable chest units), their use may also be accompanied by decreased spatial resolution and increased noise.

Analytic techniques introduced into the field by Rossmann, Doi, Morgan, and others in the early 1960's have not provided a simple predictive methodology of assessing diagnostic screen/film quality when used in a general diagnostic imaging task (1-6). On the other hand, in those cases where a particular diagnostic signal could be defined and localized, such as a vessel in angiography, bone detail in skeletal radiography, and punctate calcifications in mammography, excellent results have been obtained (7-9). One problem of more general extension is that most analytic image evaluation techniques involve application of two-dimensional optical measurements to an image of a three-dimensional object. Thus, one has a complicated perceptual task of unfolding information from superimposed planes of differing radiographic magnification (10,11). Moreover, the complicated anatomic patterns are enhanced or diminished by the use of different screen/film systems to the point where they may actually constitute "structure noise." Then, too, there is a lack of published data on physiological motion and how it enters into the final "diagnostic" evaluation.

The Quality Assurance Committee of the Department of Radiology at The George Washington University Medical Center, composed of physicians, physicists, engineers, and technologists, decided to obtain empirical data on the relationship between judgments of diagnostic quality and perceived physical attributes of radiographs, for a selection of screen/film systems. This correlation (or its lack) is of fundamental importance in the choice of an imaging system.

Members of the Quality Assurance Committee devised a test protocol to examine diagnostic and physical quality of different radiographic screen/film systems. These systems included the conventional system used in the department (Dupont High Plus/Cronex 4) and four other systems which offered relative system speeds that are a factor of two or more faster or slower than the conventional system. The investigation arose from discussions on the optimal choice of a screen/film combination for portable chest procedures, an application where system speed could be of major importance.

In this paper we discuss and illustrate methodology, realizing that much of the work is still developmental. Wherever possible, we retrospectively point out possibilities for improvement in design and/or statistical analysis. We also investigated whether an anthropomorphic, static phantom could be related to actual patient data. Moreover, we sought useful relationships between an evaluation of clinical diagnostic image quality and analytic image measurements made on the screen/film systems.

DATA COLLECTION

SAMPLE RADIOGRAPHS

Radiographs were made of patients and of the modified 3M chest phantom shown in Figure 1. An improved vasculature system and lung parenchyma were introduced into the phantom, the appearance of which was deemed reasonable by participating radiologists.

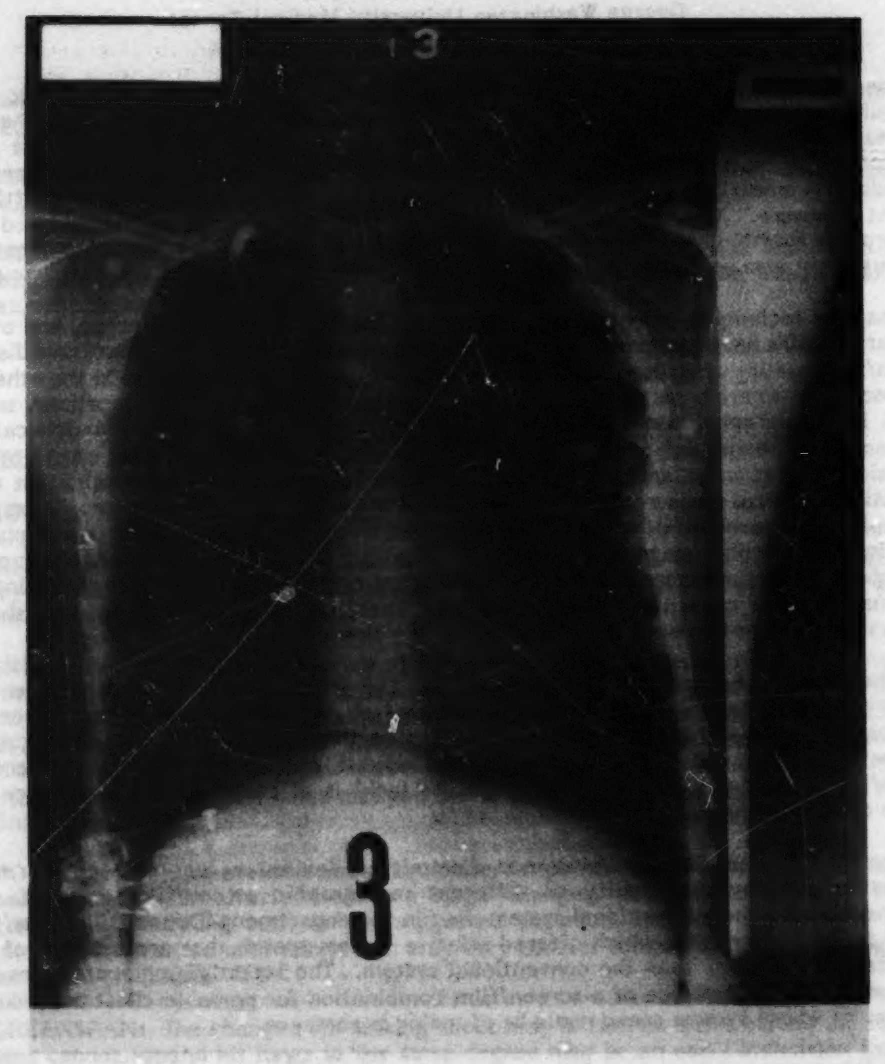


Figure 1. Radiograph of static phantom with accompanying aluminum step wedge.

The phantom was radiographed twice with each screen/film system under consideration. Each patient involved with the evaluation was radiographed using the department's conventional system plus one other screen/film system. Eight patients were involved. Only those patient radiographs judged to have equal optical density (O.D.) in the lung parenchyma were included in the evaluation. Phantom radiographs were required to differ in optical density by less than 0.10 at selected sites in the intercostal spaces. The optical density of patient radiographs was, of course, more difficult to control than that of the phantom radiographs.

In this study, because of practical considerations, one x-ray unit was used to produce the patient radiographs and another for the phantom radiographs. In order to assess any bias introduced by the use of different x-ray units, measurements were made as detailed in Table 1.

Table 1. Comparison of radiographic conditions used to produce patient and phantom radiographs

Parameter	Unit I (patient)	Unit II (phantom)
1. HVL (mm Al equivalent)	2.5	2.5
2. Peak tube potential as inferred with Wisconsin Test Cassette	80 kVcp (constant potential)	80 kVp (single phase)
*3. Effective focal spot size as measured with a star test pattern	1.55 x 1.40 mm	2.87 x 2.33 mm
*4. Focal spot - film distance	64 inches	72 inches
*5. Average midline object-film distance	59 inches	62 inches
6. Grid used	none	none

*The different focal spot sizes and different distances were somewhat compensatory and to a first order approximation resulted in equal geometric unsharpness in the midline region. At the extreme anterior and posterior regions however, the calculated geometric unsharpness differed considerably between the patient and phantom conditions. The effect of this difference is not known and warrants further consideration.

All systems were mounted in the same type of cassette and uniformprocessor performance was assured by sensitometric control. Fixed mA (mAs changed by varying time) was used so that the focal spot size would not be expected to change. System speed, screen/film characteristics and an indication of resolution appear in Table 2.

Table 2. Description of screen/film systems

System	Relative speed	Screen	Film	Resolution at a contrast response of 40 percent
A	0.55	calcium tungstate	conventional, wide latitude	1.6 l.p./mm*
B	1.00	calcium tungstate	conventional	1.6 l.p./mm*
C	1.60	rare earth	conventional	1.6 l.p./mm**
D	1.90	rare earth	green-sensitive	1.2 l.p./mm*
E	2.50	rare earth	conventional, wide latitude	1.9 l.p./mm*

*From: "Operational Health Physics," Proceedings of the Ninth Mid-Year Topical Symposium of the Health Physics Society, Denver, Colorado (1976)
**Manufacturer's data

EVALUATION OF THE RADIOGRAPHS

Radiologists were asked to choose, on the basis of their clinical experience, anatomic landmarks appearing in posterior/anterior (P/A) chest radiographs whose visualization is related to the visualization of significant radiologic findings. These anatomic landmarks were then integrated into an evaluation form, shown in Table 3. In addition, physical parameters defined and interpreted by radiologists were assembled into a second evaluation form, shown in Table 4.

Three radiologists were asked to evaluate each radiograph using only one type of form at a session. Consistent viewing conditions were maintained. The radiographs were presented to the radiologists one at a time in an arbitrary, but consistent order. The visibility of each anatomic landmark was rated by marking the appropriate box on the evaluation form. The weighting factors used to calculate a score are shown in parentheses in Tables 3 and 4, although these were not displayed on the evaluation forms actually used by the radiologists. These weighting factors reflect the relative importance placed upon each anatomic landmark or physical parameter by the radiologists.

Table 3. Subjective assessment of anatomic landmarks

(Maximum total score = 100)

Bony thorax (ribs & clavicles) (10)

Cortical margins (5)

- ( ) optimally visualized (5)
- ( ) adequately visualized (4)
- ( ) poorly visualized (2)
- ( ) not visualized (0)
- ( ) disease precludes evaluation (N)

Trabeculae (5)

- ( ) optimal detail (5)
- ( ) adequate detail (4)
- ( ) poor detail (2)
- ( ) not visualized (0)
- ( ) disease precludes evaluation (N)

Retrocardiac area (30)

Left diaphragm (15)

- ( ) totally visualized (15)
- ( ) partially visualized (7)
- ( ) not visualized (0)
- ( ) disease precludes evaluation (N)

Spine (15)

- ( ) too well visualized (7)
- ( ) optimally visualized (15)
- ( ) acceptably visualized (7)
- ( ) poorly visualized (3)
- ( ) not visualized (0)
- ( ) disease precludes evaluation (N)

Trachea (15)

Visible to: (15)

- ( ) left main stem bronchus (15)
- ( ) carina (10)
- ( ) neck and upper mediastinum (5)
- ( ) not visible (0)
- ( ) disease precludes evaluation (N)

Diaphragm (15)

Diaphragm outline (15)

- ( ) both visualized (15)
- ( ) right only (7)
- ( ) left only (7)
- ( ) not visualized (0)
- ( ) disease precludes evaluation (N)

Pulmonary vasculature (30)

Maximum measurable to: (30)

- ( ) right costophrenic angle (30)
- ( ) right mid-lung (20)
- ( ) right descending pulmonary artery (10)
- ( ) none (0)
- ( ) disease precludes evaluation (N)

DATA ANALYSIS

SCORING

The weighting factors assigned to each major parameter were divided among the response choices for each sub-parameter, as illustrated in Tables 3 and 4. The assigned point scores for the response choices were used to develop total percent possible scores for each screen/film system, for each sample radiograph, and for each radiologist. Scores were adjusted, as illustrated in Example 1, for the case in which an anatomic landmark could not be visualized because it was obscured by pathology. The percent possible scores were then averaged over the appropriate radiographs to produce a mean score for each radiologist's rating of each screen/film system.

Table 4. Subjective assessment of physical parameters

(Maximum total score = 100)

Contrast (35)

☐ optimal (35)
☐ good (23)

poor, but diagnostic
☐ too gray (11)
☐ too black/white (11)

unacceptable, not diagnostic
☐ too gray (0)
☐ too black/white (0)

Graininess (20)

☐ no grain visible (20)
☐ minimal grain (13)
☐ grainy, but does not interfere with diagnosis (6)
☐ grain interferes with diagnosis (0)

Density (15)

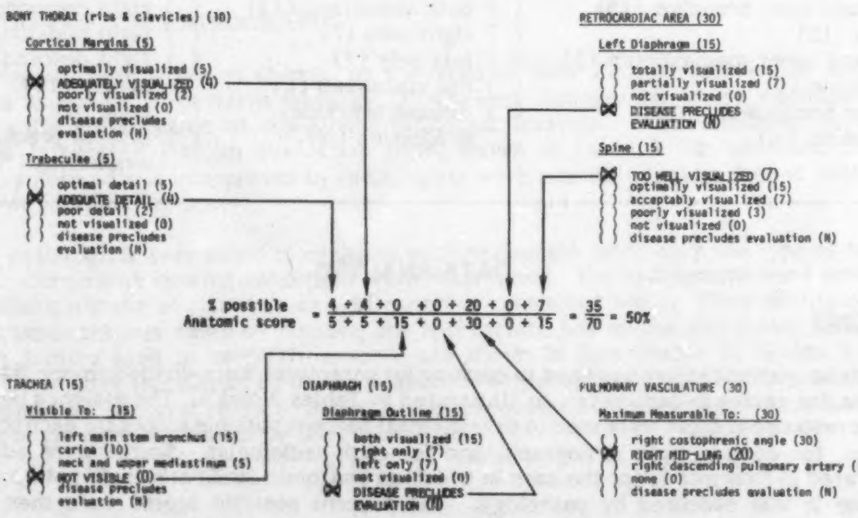
☐ optimal (15)
☐ good (10)

poor, but diagnostic
☐ too dark (5)
☐ too light (5)

☐ unacceptable (0)

Detail (30)

☐ optimal (30)
☐ good (20)
☐ poor detail, but does not interfere with diagnosis (10)
☐ lack of detail interferes with diagnosis (0)



BONY THORAX (ribs & clavicles) (10)

Cortical Margins (5)

☐ optimally visualized (5)
☒ ADEQUATELY VISUALIZED (4)
☐ poorly visualized (2)
☐ not visualized (0)
☐ disease precludes evaluation (N)

Trabeculae (5)

☐ optimal detail (5)
☒ ADEQUATE DETAIL (4)
☐ poor detail (2)
☐ not visualized (0)
☐ disease precludes evaluation (N)

RETROCARDIAC AREA (30)

Left Diaphragm (15)

☐ totally visualized (15)
☐ partially visualized (7)
☐ not visualized (0)
☒ DISEASE PRECLUDES EVALUATION (N)

Spine (15)

☒ TOO WELL VISUALIZED (7)
☐ optimally visualized (15)
☐ acceptably visualized (7)
☐ poorly visualized (3)
☐ not visualized (0)
☐ disease precludes evaluation (N)

$$\text{\% possible Anatomic score} = \frac{4+4+0+0+20+0+7}{5+5+15+0+30+0+15} = \frac{35}{70} = 50\%$$

TRACHEA (15)

Visible To: (15)

☐ left main stem bronchus (15)
☐ carina (10)
☐ neck and upper mediastinum (5)
☒ NOT VISIBLE (0)
☐ disease precludes evaluation (N)

DIAPHRAGM (15)

Diaphragm Outline (15)

☐ both visualized (15)
☐ right only (7)
☐ left only (7)
☐ not visualized (n)
☒ DISEASE PRECLUDES EVALUATION (N)

PULMONARY VASCULATURE (30)

Maximum Measurable To: (30)

☐ right costophrenic angle (30)
☒ RIGHT MID-LUNG (20)
☐ right descending pulmonary artery (10)
☐ none (0)
☐ disease precludes evaluation (N)

Example 1. Calculation of per cent possible anatomic score for one radiograph as evaluated by one radiologist

A rank was assigned to each system for each radiologist and each parameter by ordering the systems according to their mean scores. If two or more systems received an equal mean score, the average of the two ranks was assigned to both. The results for each radiologist were averaged to yield a mean rank for each system and parameter. These mean ranks were calculated separately for the scores from patient radiographs and from phantom radiographs. Example 2 shows the ranks assigned for the total of anatomic landmarks, which includes the scores for all anatomic landmarks in Table 3.

System	Radiologist I		Radiologist II		Radiologist III		All radiologists
	Mean score	Rank	Mean score	Rank	Mean score	Rank	Mean rank
A	76.4	(2)	83.6	(2)	79.0	(2)	(2.0)
B	74.6	(3)	73.7	(3)	65.6	(3)	(3.0)
C	60.1	(4)	70.6	(4)	43.5	(5)	(4.3)
D	94.3	(1)	85.8	(1)	82.0	(1)	(1.0)
E	49.6	(5)	56.1	(5)	60.0	(4)	(4.7)

Example 2. Mean scores for total of anatomic landmarks (for patient radiographs) with ranks assigned

Mean ranks as a function of relative speed are displayed in Figures 2, 3, and 4. Note that the patient data curve for the total of anatomic landmarks (Figure 2) displays the mean ranks calculated in Example 2.

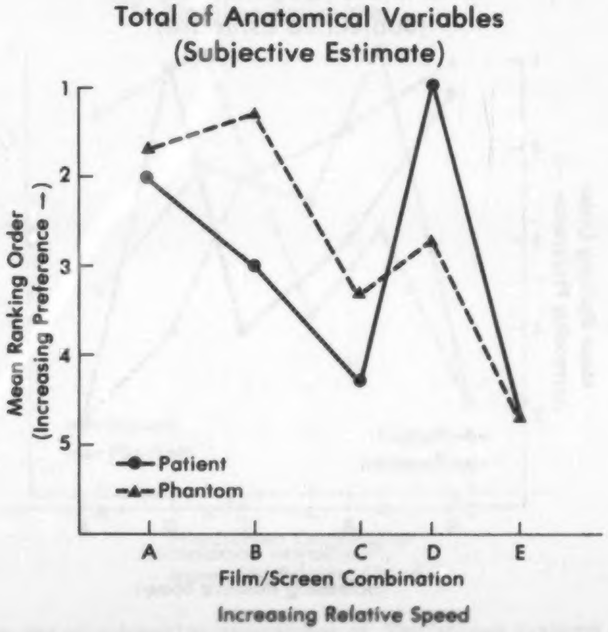


Figure 2. Results of subjective assessment of film/screen systems using anatomic variables shown in Table 3.

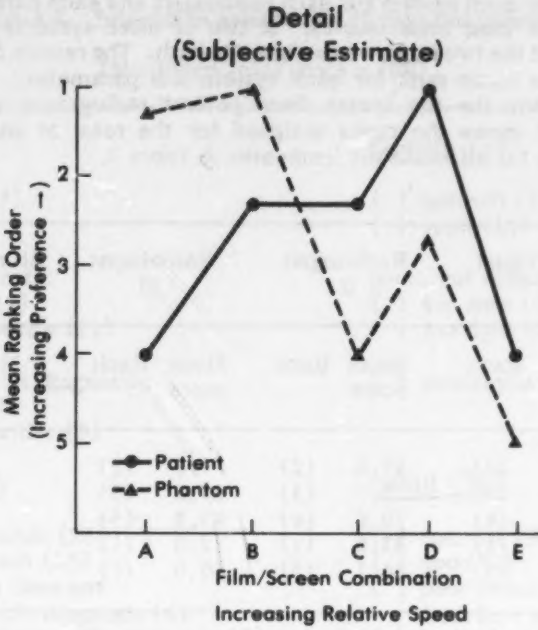


Figure 3a. Results of subjective assessment of film/screen systems using only the scores from the detail category of Table 4.

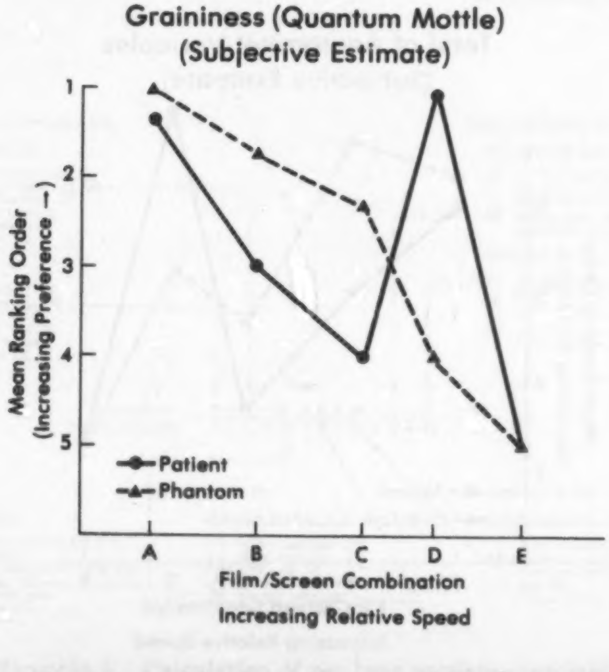


Figure 3b. Results of subjective assessment of film/screen systems using only the scores from the graininess category of Table 4.

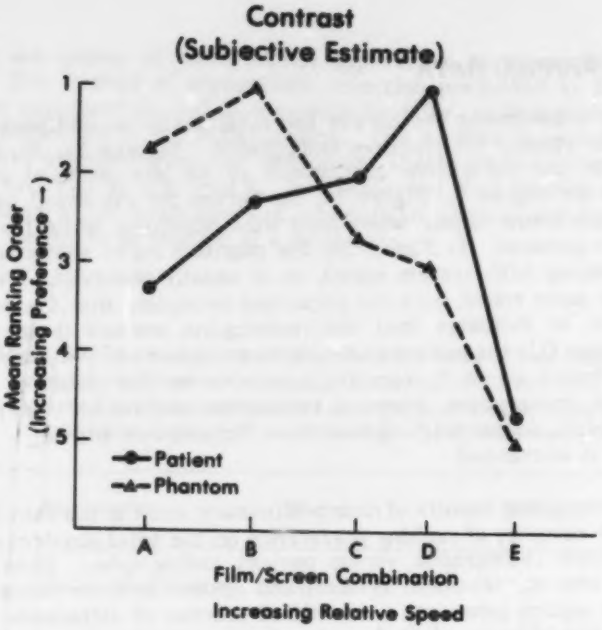


Figure 3c. Results of subjective assessment of film/screen systems using only the scores from the contrast category of Table 4.

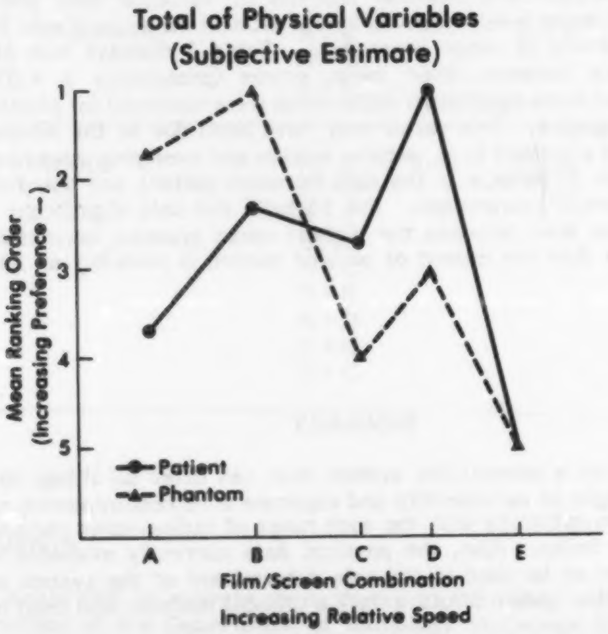


Figure 4. Results of subjective assessment of film/screen systems combining scores from all four categories of physical variables shown in Table 4.

DISCUSSION OF THE RANKED DATA

In this preliminary experiment the curves for total anatomic and physical parameters (Figures 2 and 4) agree closely for phantom radiographs. Figures 3a, 3b, and 3c show the ranking preference for the subjective assessment of certain physical parameters that were the constituents for Figure 4. Figure 3a, the curves for the detail parameter, shows that slow speed systems score higher when used with phantoms while fast systems score better when used with patients. In Figure 3b, the phantom curve shows perceived system quantum mottle increasing with system speed, as is usually observed. The patient curve in Figure 3b shows the same trend, with the important exception that System D was ranked highest (best). Figure 3c indicates that the radiologists seemed to show a preference for the contrast of System D in the patient radiographs as opposed to the phantom radiographs (Figure 3c). The performance of System D appears to be the result of optimizing the combination of motion unsharpness, intrinsic resolution, and noise. The data should be considered as "hypothesis suggesting" rather than "hypothesis proving"; however, it is felt that further study is warranted.

One of the more interesting results of this preliminary work is the fact that one finds a statistically significant reversal of ranking preference on the total physical score (Figure 4) when comparing phantom radiographs versus patient radiographs. Note especially the ranking for System B (the conventional system) and System D (over twice as fast). It is concluded that motion and/or other yet unidentified sources of difference between static phantoms and patients should be carefully kept in mind in attempted evaluation schemes.

ANALYSIS OF DATA FOR SIGNIFICANT DIFFERENCES AMONG MEAN SCORES

The statistical analysis used one-way analysis of variance with unequal cell sizes at the 99 percent confidence level, followed by the Student-Newman-Keuls Test to compare the statistical significance of mean score data. Table 5 displays sets of systems with a significant difference between their mean scores (probability $p < .05$). Inspection of Table 5 indicates that more significant differences were observed for phantom radiographs than for patient radiographs. This result may have been due to the inherent differences between a phantom and a patient (e.g., patient motion and overlying structure) or variations between patients. The difference in the data between patient and phantom data is most noticeable for the "detail" parameter. For "detail" the only significant difference for patient radiographs was seen between the highest speed systems, consistent with but not proving the hypothesis that the extent of patient motion is probably an important aspect of perceived detail.

SUMMARY

There probably is not a screen/film system that can offer all things to all diagnostic tasks, particularly in light of variable kVp and exposure time requirements; nor is there one system that can perform optimally with the wide range of radiographic conditions present in radiology departments today. Also, the physical data currently available on screen/film systems are insufficient to be used as the sole determinant of the system of choice. The selection of a screen/film system is ultimately a clinical decision and must be made by the radiologists.

The authors are aware of more formal approaches to screen/film selection, such as ROC analysis. The method of screen/film selection presented in this paper is suggested not as a counter argument to, nor as a replacement for, such approaches, but as a method more easily implemented in a clinical setting. The method presented here removes some of the subjectivity inherent when the radiologist simply chooses one system over another, and also allows analysis of the results. It is expected that further application of the use of anatomic landmarks for evaluations of screen/film systems will lead to better definition of the potential signals to be studied at a later date.

Table 5. Pairs of screen/film combinations whose means of a specified parameter are significantly different at $p < .05$ (listed in decreasing order of significance; screen/film combination on left had the higher score of each pair)

Parameter	Phantom	Patient
Total Anatomical	B > E A > E	D > E A > E D > C
Contrast	B > E A > E	D > E
Graininess	A > E A > D B > E	D > E
Detail	B > E B > C A > E B > D A > C D > E A > D C > E D > C	D > E

ACKNOWLEDGMENTS

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A MODULAR CHEST PHANTOM

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There are many reasons why one might want to build a chest phantom. Most of these reasons will fit into one of the following three categories:

1. Technical: to assist in such tasks as standardizing radiographic techniques and comparing various combinations of screens, films and kilovoltages.
2. Educational: as a "patient" for beginning technologists to practice techniques.
3. Investigative: as a research tool to examine both imaging techniques and reader performance.

Before beginning construction, it is imperative that the reason for building the phantom be precisely defined. A phantom may be a very effective tool for technical evaluations and not resemble a human in any way. An example of such a phantom is the one designed by Littleton to evaluate tomographic systems (1). This phantom consists of a combination of wire screens and rods cleverly oriented to assess various parameters of tomographic performance. A phantom for education, e.g., for technician training, should look as much like a human as possible. The phantom itself should resemble a human and so should its radiographic image. Phantoms built for research purposes should be designed to meet the needs of the specific project.

The characteristics of an ideal phantom for research are as follows:

1. X rays of the phantom should look exactly like those of a human chest and contain all of the components of the thorax and upper abdomen, including the lungs, mediastinum, heart, chest wall and diagphragm.
2. The phantom should have all tissue-equivalent components so that contrast remains "true" at various beam energies.
3. The phantom should have interchangeable components, so that readers will not be able to learn its anatomic nuances.
4. The phantom should accept disease in a physiologic fashion, i.e., pulmonary nodules should be incorporated into the substance of the lung rather than superimposed on the image of the lungs.
5. The phantom should accept both nodular and infiltrative disease processes.
6. Structural and temporal stability are other important features.

The construction of an ideal phantom is exceedingly difficult, if not impossible. Almost always, a compromise must be made. The nature and degree of compromise depend upon the design of the experiment. Our modular chest phantom (Fig. 1) is very much a compromise. Our initial aim was to investigate the effect of quantum mottle on the perception of pulmonary disease. We reasoned that mottle would affect nodule detection more than infiltrate detection because of the smaller size of the nodules. We further reasoned that the eye perused only a relatively small portion of a chest film at any one instant when searching for a nodule. Therefore, we did not need a whole chest phantom to accomplish our objective. We settled on the left lung base of a fairly young person (Fig. 2A).

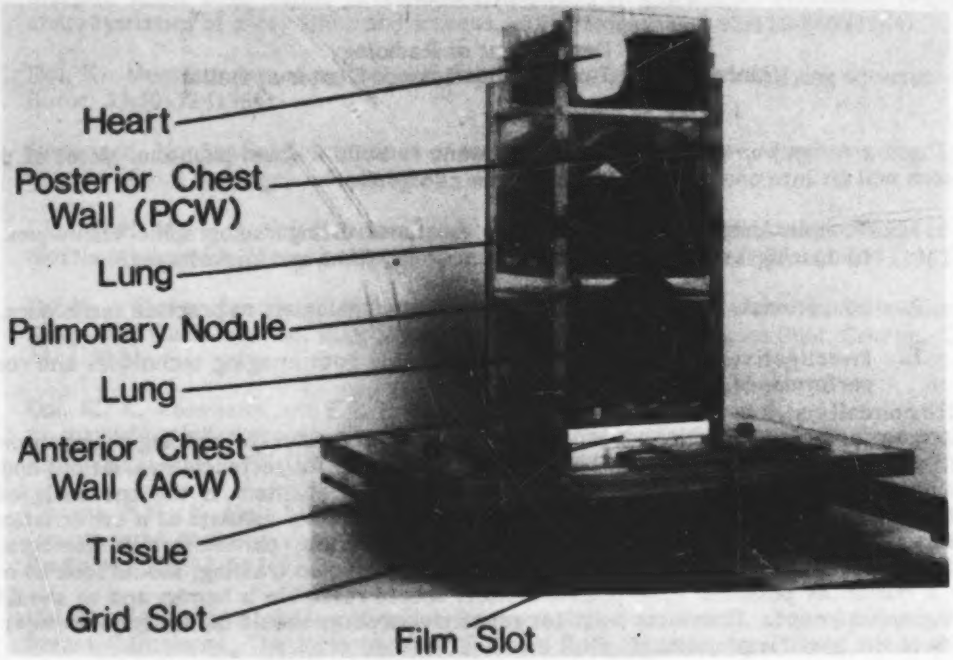

Figure 1. Photograph of the modular chest phantom.

CONSTRUCTION OF THE PHANTOM

Each component of the phantom is contained in a separate drawer, or module (Fig. 1). The modules are then superimposed in a 1 cm thick Lucite tower. The components are assembled to simulate a postero-anterior chest film with the posterior ribs close to the x-ray tube. The components are assembled as follows beginning at the top of the phantom:

1. Heart - "tissue equivalent" wax 5 cm thick, molded to the contour of the left heart border and held in place with a low density polyurethane sponge.

2. Posterior ribs - human ribs cut to size and imbedded in household paraffin 2.5 cm thick, molded into the contour of the posterior chest wall.

3. Lungs - one lung of a large mongrel dog, inflated by a vacuum technique, fixed with alcohol and formaldehyde vapor for 48 hours, injected with silicone rubber (RTV 602, specific gravity 0.99) and cut to size under fluoroscopy.

4. Nodule tray - a 1/8 in thick lucite plate with 1 cm of clearance above and 0.3 cm below. Beeswax spheres 3-7 mm in size were used as test objects. The nodules were placed in low density polyurethane sponge to simulate the lungs.

5. Lung - opposite lung of the same dog, prepared in the manner described above.

6. Anterior ribs - appropriate anterior human ribs prepared the same as the posterior ribs but in 1.5 cm thick paraffin.

7. "Obesity compartment" - a space to add up to 3 cm of tissue equivalent wax to simulate an obese patient.

8. Lead mask - 1/8 in thick lead sheet with a 10 x 12.5 cm hole. The lead blocked out the image of the housing.

9. Film (8 x 10 in) and Buckey slots (10:1 stationary grid focused to 60 in). Films were usually exposed to a density of 1.30 ± 0.05 over a selected open area of the lung.

10. Numbering slot - for film identification (from 0-999).

EXPERIMENTAL DESIGN

All studies were done with from 0-4 nodules (average 3) per film. Usually 40 films were exposed for each technique. If eight techniques were being compared, a total of 320 films (40 x 8) would be used. An approximately equal number of 4, 5, 6 and occasionally 7 mm nodules were randomly distributed upon the nodule tray (avoiding the hilum because these nodules could not be detected). The films were numbered with the following scheme:

"Randomization" of Numbers
(3 techniques - 40 films/technique)

Nodule Placement	Technique		
	#1	#2	#3
1	1	41	81
2	82	2	42
3	43	83	3
4	4	44	84

After nodules were placed on the tray, their centers were marked for future localization with tiny lead shot taped to the bottom of the tray. After making an exposure with the nodule positions marked, the shot was removed and the test films were exposed, one for each technique. These films were numbered as shown above, i.e., 1, 41 and 81. After each nodule placement (3 films), the numbering sequence shifted to the right. In this way, the film reader saw all three techniques over-and-over, but he had to read 40 films before he saw another film with a duplicate nodule placement.

FILM INTERPRETATION

The films were interpreted in a specially designed holder which demanded rather precise localization. The holder was divided into 1/8 in increments along the x- and y-axes (20 on the x-axis and 25 on the y-axis). To be counted as a true positive, a nodule had to be localized to plus or minus one increment, i.e., to one of nine points out of a total of 500 (20 x 25) possible points. The odds were 56:1 against a "chance" localization (500 ÷ 9 = 56).

In the first few studies that we did with the film holder, we gave the reader an unlimited amount of viewing time. Some of our residents took 3-4 hours to interpret 50 films. Fatigue became a problem--not to performance, but to "recruiting" volunteers. This problem was overcome with a 20-second clock (like the 24-second clock used in professional basketball). The viewbox was plugged into a dark room timer that turned the light off at 20 seconds. With this technique, a 120-film study could easily be read in one hour:

Read time: 20 seconds/film x 120 films = 2400 seconds
Change time: 5 seconds/film x 120 films = 600 seconds
3000 seconds

3000 ÷ 60 = 50 minutes

The reader called out his localizations and they were recorded by a technician. All studies were interpreted by 10 readers. Initially, we only used senior residents and staff radiologists, but the results were exactly the same no matter how well the reader performed. For example, a staff radiologist usually saw more nodules than a technician, but the qualitative results were the same. If technique A was better for the radiologist, it was also better for the technician. For some of the later studies, half of the readings were done by technologists.

STUDIES PERFORMED WITH THE MODULAR PHANTOM

Five studies have been performed with the modular phantom:

1. Low kilovoltage (70 kVp) non-grid versus intermediate energy (120 kVp) grid techniques (2). Grid techniques proved superior to the non-grid technique for all nodule sizes for every single reader. The average number of nodules detected (all readers and all nodule sizes) was 42 percent with the 70 kVp non-grid technique and 55 percent with the 120 kVp grid technique. These results show the detrimental effect that scatter radiation has on image quality and point out the necessity of using a grid, air gap or moving slot technique.

2. The effect of film density on nodule detectability (unpublished). Films were exposed to densities of .75, 1.0, 1.25 and 1.5 ± 0.05 over a selected area of the open lung, i.e., lung not overlaid by rib. Detection rates were virtually identical with all four techniques.

3. Intermediate (100 and 150 kVp) versus high (200 and 300 kVp) energy techniques (3) This study was done with a General Electric Maxitron 300 radiation therapy unit at a target-to-film distance of 9.8 meters to decrease the "effective" focal spot size from 5.6 mm to approximately 1.0 mm. Nodule detection rates increased with increasing beam energy up to 200 kVp, and then decreased slightly at 300 kVp. Surface exposures decreased up to 200 kVp and increased slightly at 300 kVp. Conclusion: The optimum energy for the detection of pulmonary nodules appears to be approximately 200 kVp. Note: The first half value layer of our 200 kVp beam is 1.3 mm of copper, which is almost exactly the same as the 1.2 mm of copper of the Hewlett-Packard 350 kVp unit.

4. The effect of fatigue on resident performance (4). Ten residents read a series of 25 films on two occasions at least two months apart. One reading was done early in the day and another after at least 14 hours of continuous work. Half did the first readings rested and half fatigued. The results of the two readings were identical.

5. The effect of structured noise (5) on nodule detectability (6). This study was done to evaluate the relative importance of the ribs and lung vasculature in obscuring pulmonary nodules. It is the type of study that can only be performed with a nodular phantom. A series of eight radiographs was exposed with the phantom in various stages of disassembly. These eight films were done in two sets of four, one set at 70 kVp and the other at 120 kVp.

Figure 2 shows the four films included in each set: (A) the complete phantom, (B) the ribs removed, (C) the lungs removed and (D) the ribs and lungs removed. The results of the study are as follows (average of ten readers, 320 films, 960 nodules):

	Percent of true positives	
	70 kVp	120 kVp
Whole phantom	31	42
Ribs removed	62	65
Lungs removed	74	82
Ribs and lungs removed	97	95

These data show several important points: First, they point out the superiority of higher energy techniques for showing pulmonary nodules. Second, they show that the lungs are more detrimental to nodule detection than the ribs. Third, almost all nodules in the 4-6 mm range are detected after removal of the lungs and ribs. Fourth, when the ribs are removed, the 70 kVp and 120 kVp techniques are virtually identical. Differences of 3 percent, or less, have little statistical significance. However, the most important finding in the study is that structured noise sources do not act independently. If they independently behaved, for example, like the transmittance of light through a film, then their combined effect would simply be the product of their individual effects. If light transmittance is 40 percent through one film and 80 percent through another, the transmittance of the two films superimposed is 32 percent (.40 x .80 x 100). At 70 kVp, the nodule detection rates are 62 percent with the ribs removed and 74 percent with the lungs removed. Ignoring the relatively unimportant third factor which prevents 100 percent detection with both removed, the combined effect if the two acted independently would be 46 percent (.62 x .74 x 100). However, the experimentally measured rate was only 31 percent, a statistically significant difference. The combined effect is much greater than might have been anticipated. Therefore, even a relatively minor decrease in the importance of one structured noise source, like using a higher kVp to decrease the visibility of ribs, may have a fairly major effect on nodule detectability.

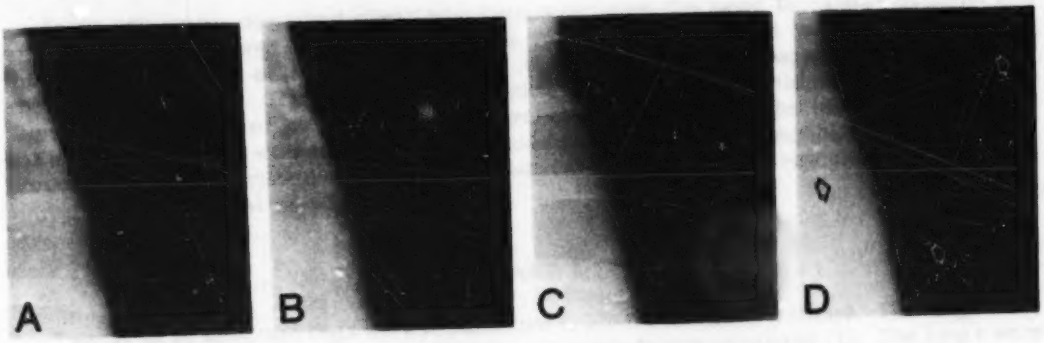

Figure 2. Radiographs of the modular chest phantom in various stages of disassembly.

SUMMARY

1. Scatter radiation is always detrimental to image quality. A grid, or air gap or moving slot is essential.

2. Increasing beam energy is advantageous, for pulmonary lesions, up to approximately 200 kVp. Higher energies offer little, if any, advantages and have the disadvantage of increasing patient exposure.

3. Multiple structured noise sources interrelate in a complex manner, but the net effect is greater than would be anticipated from the various sources acting independently.

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DEVELOPMENT OF A RADIOGRAPHIC CHEST PHANTOM WITH DISEASE SIMULATION

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INTRODUCTION

The evaluation and quality assurance of diagnostic x-ray systems are facilitated by the use of phantoms that have anthropomorphic geometries (1,2). This is particularly true for chest radiography, which accounts for approximately 40 percent of radiographic examinations and often involves multiple radiographic techniques.

Radiological personnel, particularly radiological technologists, engineers and diagnostic physicists have in the past relied on test objects such as resolution charts, step wedges, Lucite beads, and steel wool in an attempt to measure total performance of their radiographic systems (3). Although some quantitative information may be derived from such tests (e.g., resolution or MTF estimates) these test objects have substantial disadvantage because the radiographic image produced bears no resemblance to the radiograph of the human body. Patients cannot be used for safe, detailed comparative evaluation of diagnostic techniques because of radiation exposure and the variation of anatomy from patient to patient. The previous test objects are unable to simulate disease processes and this creates limitations when they are the only tool used to attempt to characterize a complex diagnostic entity such as a chest radiograph. There are no detailed or controlled comparisons available to compare the "language" of the physicists with the "language" of the diagnostic radiologist.

Some previously described chest phantoms overcame the need to expose patients and allowed the positioning of test objects (4,5). Other designs which had reasonable intrathoracic detail were either too difficult to make (6), or did not represent actual human chest dimensions (7). The major problem with the presently available human thorax phantoms is the absence of the vascular and parenchymal detail and the lack of the variable pathological patterns common to clinical radiology. This paper describes a technique for producing a chest phantom with normal anatomical detail, including vascular structures and simulated pathological conditions.

MATERIALS AND METHODS

Twenty to 30 kilogram dogs were heparinized (1000 units) and sacrificed. The lungs were carefully removed and the trachea and main pulmonary artery canalized (1). The lungs were inflated by bubbling air through 36 percent formalin after the method of Blumenthal and Boren (8).

After initial lung inflation, the pulmonary vascular system was flushed with normal saline until the residual blood was cleared. The vascular system was then injected by hand with latex rubber until the subpleural vessels were visible.

The mediastinal insert and diaphragm were cast from tissue-equivalent rubber. The assembled lung and mediastinum were then inserted into the available thoracic cages and surrounded with urethane foam. This provided for easy handling and the option of multiple interchangeable inserts allowing for disease simulations for any one phantom model. A variety of simulated pathological conditions can be created by using urethane materials and other tissue-equivalent polymers (1). This provides for future quality control programs, prospective imaging studies, and optimization of equipment (2).

DISCUSSION

Recent Federal guidelines for radiation safety in Federal agencies specify maximum exposure limits for certain radiological examinations (2,9). However, these guidelines leave the criteria of image quality up to the institution concerned. For example, the maximum entrance skin exposure for a P.A. chest (23 cm thick) is set at 30 mR. It is possible to comply with this Federal guideline by using a fast screen-film combination with high kVp. The measured quality of the image obtained tends to be mandated by physics parameters alone with these methods and such guidelines tend not to emphasize critical subjective image quality evaluations. Radiologists, however, rarely make decisions on film-screen combinations or departmental technique policies on the basis of physics parameters alone. They require subjective assessment. This phantom provides a controlled means to produce comparative images. It allows less patient variability and more control over these studies.

Subjective image interpretation also depends on the human factor. Radiologists, in common with all physicians, have a relatively high rate of "missed diagnoses" (10,11). While part of this misinterpretation is certainly due to technical factors, i.e., the inability of the standard diagnostic radiologic systems to demonstrate lesions of low radiodensity, the significant human factor, currently being assessed by workers in the field of perception and psycho-pictorics, warrants further evaluation. Prospective analysis of both technical and human components of this error rate are now more accurately obtainable with this three-dimensional chest phantom. Thus, the ability to use this phantom, in either its "normal anatomy" mode or with its variable disease simulation, will assist radiologists to reach a more definitive understanding of the information contained in a chest radiograph, and asist them to understand their abilities to arrive at the proper diagnosis.

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THE USE OF AN ANTHROPOMORPHIC CHEST PHANTOM

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INTRODUCTION

New radiographic imaging techniques, systems and ideas are constantly being introduced to the field of radiology. The obvious question raised is whether a new imaging system provides more information than before or, alternatively, the same information at a lower patient dose. A radiologist does not generally use numerical values of noise, resolution and contrast to determine whether a particular imaging system is suited to his needs. Instead he uses his own subjective impressions of these parameters. In order to make an evaluation, a radiologist may try a new technique or system by actually using patients under clinical conditions, or he may use test objects such as human bone phantoms. Patients differ in their size, shape and internal structure which makes it difficult to perform controlled studies. This, combined with the clinically unproductive radiation they may receive, makes them unsuitable for use as test objects in radiography. Bone phantoms are ideal for predicting how well a new technique will image bone, but they are not much use for predicting how well the soft tissue structures of the chest will be imaged.

A chest phantom employing existing thorax phantoms and inflated dog lungs injected with blood-equivalent rubber and simulated lung pathology has been developed (1). Such a phantom serves as a standard patient to bring a greater degree of objectivity to the evaluation of new techniques using data relevant to the everyday practice of radiology. Some of the ways in which this phantom has been used are discussed and suggestions for other uses are given. (NOTE: The chest phantom referred to in this study measures 23 cm from front to back, and is radiographically equivalent to a 165-pound normal male of the same thickness.)

RADIOGRAPHIC TECHNIQUE SELECTION

The principal use of this phantom has been to provide a clinically relevant method to compare images produced by different radiographic techniques used in diagnostic radiology. In preliminary experiments we have studied the effects on the information content of chest radiographs as a result of varying (a) kVp, (b) screen/film combination, and (c) screen/film combination and kVp. The results of these experiments are still being analyzed and experimentation is continuing. A brief description of each experiment and the techniques used is given here.

kVp

The phantom, cassette holder and x-ray tube were rigidly mounted on a steel beam to eliminate motion artifacts. The focal film distance was 72 in. Water density rubber nodules of varying sizes (> 3 mm) were placed at several positions outside the phantom in front of the grid. The locations of the nodules were chosen such that they would be projected behind ribs, within the mediastinum, superimposed on vessels and close to the lateral chest wall. Radiographs were made at the full range of kVp values available, from 40 to 150 kVp. Du Pont Hi Plus screens were used with Kodak RP film in an Xomatic cassette. The processor was monitored sensitometrically and maintained at a constant speed point. The density of each film was matched by changing mAs. The techniques used and entrance exposures are given in Table 1.

Table 1. Techniques used to produce a series of nodule phantom radiographs with varying kVp.

Film No.	kVp	mAS	ESE[a]	Filter	AFD	OFD[b]	Grid[c]
0	40	139	59	1 mm Al	200 cm	2 cm	no
1	50	25	20	1 mm Al	200 cm	2 cm	no
2	60	10	12	1 mm Al	200 cm	2 cm	no
3	70	5	9	1 mm Al	200 cm	2 cm	no
4	80	3	7	1 mm Al	200 cm	2 cm	no
5	90	2	5	1 mm Al	200 cm	2 cm	no
6	100	1.3	4	1 mm Al	200 cm	2 cm	no
7	125	.6	4	1 mm Al	200 cm	2 cm	no
8	80	17	25	1 mm Al	200 cm	2 cm	yes
9	90	10	19	1 mm Al	200 cm	2 cm	yes
10	100	7	16	1 mm Al	200 cm	2 cm	yes
11	125	3	14	1 mm Al	200 cm	2 cm	yes
12	150	1.3	5	1 mm Al	200 cm	2 cm	yes
13	150	4	1.5	1 mm Cu	200 cm	2 cm	yes

a. Entrance Skin Exposure (ESE) measured in air at position where posterior chest wall would be (PA projection), MDH model, 1015C, 6 cc chamber.

b. Anterior chest wall-to-film distance

c. Leibel-Flarsheim, 10:1, 103 line, Al interspaced, 60-72" focussed grid

One shortcoming of this series of films is the fact that the nodules were located outside the chest, close to the film. This prevents them from being undercut by scattered radiation. This effect is very noticeable at low kVp values but decreases in importance as the energy of the beam increases, because there is more penetration of the nodules by the primary beam at higher kVp values. Therefore the effect on image contrast is the same. This is a good example of what not to do when using a phantom to study the effects of kVp on image quality. It is important to have the simulated lesions placed in the same anatomical position as they would be in a patient. Further work is planned in this area.

SCREENS AND FILMS

Seventy-two screen/film combinations from four major manufacturers were used in this study. Each screen was mounted in a Kodak Xomatic cassette. The chest phantom containing simulated pulmonary nodules greater than 3 mm and other simulated pathology was set up on an x-ray "optical" bench. The tube voltage remained constant at 100 kVp and the exposure changes were made by varying mAs. Film processing (Kodak SP, Gevamatic chemistry, 86 °F, 3'20" cycle) was monitored sensitometrically and the radiographs were exposed to a density of 1.2 ± 0.1 measured at the same point on the film. Table 2 lists the combinations and the exposure factors used. The entrance skin exposure (ESE) was measured in air at a point corresponding to the back of the chest (PA projection) and normalized to a density of 1.2 for all combinations. The speeds of the systems varied by a factor of 18 from the slowest (91 mR) to the fastest (5 mR). These results have been used to select screen/film systems for use in our medical center, as well as to serve as a basis for other experiments.

From this study it became apparent that it is difficult to get a consensus from several radiologists grading the image quality of a large number of radiographs, especially when many of them are close in quality. It was difficult to grade the films using the criterion of "diagnostic" quality since in most of the radiographs all of the simulated pathology was detected. Therefore the films were considered diagnostic for the task of nodule detection. Once a radiologist detected a nodule on a good quality radiograph, he was able to detect it on a poorer quality film since he already knew where to look. Therefore the radiologist observers relied somewhat on their own personal interpretation of image quality to grade the films. Criteria such as base tint and preferences learned during residency came into play during the selection process. For example, some observers preferred "high contrast" over "latitude" films for no reason other than that was what they were used to looking at. Also, intangibles such as "sparkle" and "snap" were found to play a role. Thus, most of the radiographs were judged on their "picture" quality or eye appeal rather than diagnostic quality. However, if the radiologists were supplied with more information on the films of equal quality, such as entrance exposure, the selection process became much easier.

A major disadvantage of this type of phantom is that the observer quickly learns the locations of rib crossings, end-on blood vessels and other structured noise. This is not representative of the clinical situation. Several phantoms would be needed if they were to be used to make radiographs for observer performance studies.

SCREEN/FILM AND kVp

In another experiment we selected 16 screen/film combinations and three kVp levels, and produced a series of 48 chest phantom radiographs. The kVp settings were 80, 115, and 150 using a 10:1, 103 line grid. The screens were chosen such that a wide range of phosphor types was represented (Table 3). The aim was to compare the image quality of chest radiographs over a wide range of exposure values in order to find out what quality of radiograph was obtainable within the recommended guidelines of 30 mR ESE (2). Again, the density of each film was carefully matched in the lung fields and the processing strictly controlled. The exposure to each film was varied by changing the mA and time. Because the exposure times varied over a wide range, the possibility of reciprocity law failure must be considered when comparing the relative speeds of different systems.

A panel of radiologists viewed the films and arrived at a subjective impression of image quality, with special reference to the task of nodule detection. The radiographs were grouped into three categories: most acceptable, acceptable, and least acceptable. Inevitably, there was some overlap using this type of grading system, but the general trend was a deterioration in image quality as exposure decreased. There were a few notable exceptions to this trend, however, one of which is shown in Figure 1. These two films were judged to be of equal image quality but differed in entrance exposure by approximately 300 percent.

Table 2. Comparison of 72 film-screen combinations using the anthropomorphic chest phantom. All films were matched in density + 0.1. Normalized E.S.E. = entrance skin exposures normalized to a density of 1.2. Processing was in Agfa "Gevamatic" chemistry, 40 sec. immersion time, 86° F.

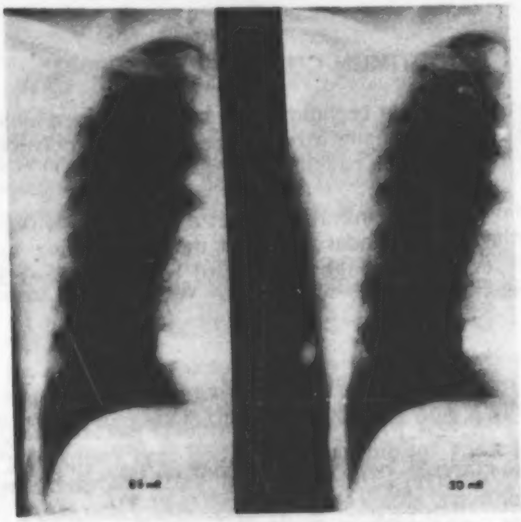
SCREEN	FILM	kVp	mAS	Normalized ESE	SCREEN	FILM	kVp	mAS	Normalized ESE (mR)
PAR	CR4	100	11	55	Quanta II	HL	100	2.4	13
PAR	CR6	100	9	38	Quanta II	R	100	2.5	12
PAR	CR6+	100	12	56	Quanta II	RP1	100	1.88	8
PAR	CR7	100	25	91	Quanta II	RP1L	100	1.6	8
PAR	HL	100	10	40	Quanta III	CR4	100	1.88	7
PAR	R	100	11	46	Quanta III	CR6	100	1.5	7
PAR	RP1	100	9	32	Quanta III	CR6+	100	1.88	10
PAR	RP1L	100	7.5	38	Quanta III	CR7	100	3.2	16
Hi Plus	CR4	100	5	21	Quanta III	HL	100	1.5	6
Hi Plus	CR6	100	4	16	Quanta III	R	100	1.5	7
Hi Plus	CR6+	100	5	26	Quanta III	RP1	100	1.3	5
Hi Plus	CR7	100	8	35	Quanta III	RP1L	100	1.0	6
Hi Plus	HL	100	4	18	XOM REG	RP	100	6	33
Hi Plus	R	100	4.5	18	XOM REG	XG	100	10	48
Hi Plus	RP1	100	3.75	15	XOM REG	R	100	6	29
Hi Plus	RP1L	100	3	15	XOM REG	HL	100	6.4	35
Hi Speed	CR4	100	7.5	31	XOM REG	RP1	100	5	21
Hi Speed	CR6	100	6	23	XOM REG	RP1L	100	3.75	15
Hi Speed	CR6+	100	7.5	36	LAN FINE	OG	100	6.4	36
Hi Speed	CR7	100	16	66	LAN REG	OG	100	1.5	8
Hi Speed	HL	100	6	29	Alpha 4	XD	100	3.2	20
Hi Speed	R	100	6.4	38	Alpha 4	XDL	100	3.75	18
Hi Speed	RP1	100	5	23	Alpha 4	XM	100	1.6	8
Hi Speed	RP1L	100	5	19	Alpha 4	XUD	100	8	36
Lightning+	CR4	100	3.75	16	Alpha 48	XD	100	3	14
Lightning+	CR6	100	3	14	Alpha 48	XDL	100	3.2	14
Lightning+	CR6+	100	4	18	Alpha 48	XM	100	1.88	8
Lightning+	CR7	100	7.5	32	Alpha 48	XUD	100	6.4	29
Lightning+	HL	100	3.2	18	Alpha 8	XD	100	2	13
Lightning+	R	100	3.5	17	Alpha 8	XDL	100	2.4	13
Lightning+	RP1	100	3	11	Alpha 8	XM	100	1.6	7
Lightning+	RP1L	100	2.5	12	Alpha 8	XUD	100	6	29
Quanta II	CR4	100	2.5	12	Alpha 12	XD	100	1	7
Quanta II	CR6	100	2	10	Alpha 12	XDL	100	1.2	7
Quanta II	CR6+	100	3	14	Alpha 12	XM	100	.75	5
Quanta II	CR7	100	5	35	Alpha 12	XUD	100	3	14

Table 3. Varying screen film combination and kVp

SCREEN	FILM	kVp	mA	S	mAS	mR	SCREEN	FILM	kVp	mA	S	mAS	mR
Dupont Par	Cr 6+	80	20	1.3	26	65	Lan Fine	OG	80	100	.2	20	50
Dupont Par (non-grid)	Cr 6+	80	100	.10	10	25	Lan Fine	OG	115	100	.05	5	28
Dupont Par	Cr 6+	115	100	.06	6	33	Lan Fine	OG	150	100	.02	2	21
Dupont Par	Cr 6+	150	100	.03	3	30	Lan Reg	OG	80	100	.05	5	12
Hi Plus	Cr 6+	80	100	.12	12	30	Lan Reg	OG	115	100	.01	1	7
Hi Plus	Cr 6+	115	100	.03	3	17	Lan Reg	OG	150	100	.002	.2	6
Hi Plus	Cr 6+	150	100	.01	1.0	12	Alpha 4	XD	80	100	.10	10	25
Quanta II	Cr 6+	80	100	.06	6	15	Alpha 4	XD	115	100	.025	2.5	15
Quanta II	Cr 6+	115	100	.016	1.6	10	Alpha 4	XD	150	100	.8	.008	11
Quanta II	Cr 6+	150	100	.006	.6	9	Alpha 4	XUD	80	100	.25	25	60
Quanta II	Cr 7	80	100	.12	12	30	Alpha 4	XUD	115	50	.12	6	32
Quanta II	Cr 7	115	100	.03	3	17	Alpha 4	XUD	150	100	.02	2	21
Quanta II	Cr 7	150	100	.01	1	13	Alpha 8	XD	80	100	.06	6	15
Quanta III	Cr 6+	80	100	.04	4	10	Alpha 8	XD	115	100	.013	1.3	8
Quanta III	Cr 6+	115	100	.01	1	7	Alpha 8	XD	150	100	.005	.5	8
Quanta III	Cr 6+	150	100	.003	.3	7	Alpha 8	XUD	80	100	.16	16	40
Quanta III	Cr 7	80	75	.1	7.5	18	Alpha 8	XUD	115	100	.05	5	28
Quanta III	Cr 7	115	75	.025	1.88	11	Alpha 8	XUD	150	100	.016	1.6	18
Quanta III	Cr 7	150	50	.013	.65	9	Alpha 12	XD	80	100	.03	3	8
Dupont SP	Cr 6+	80	100	.16	16	40	Alpha 12	XD	115	75	.01	.75	5
Dupont SP	Cr 6+	115	100	.04	4	22	Alpha 12	XD	150	50	.003	.15	5
Dupont SP	Cr 6+	150	100	.016	1.6	18	Alpha 12	XUD	80	100	.08	8	20
XOM Reg	RP	80	100	.12	12	30	Alpha 12	XUD	115	50	.04	2	11
XOM Reg	RP	115	75	.05	3.75	20	Alpha 12	XUD	150	100	.006	.6	9
XOM Reg	RP	150	75	.02	1.5	15							

Processing: Kodak RP chemistry 40 sec. dev. immersion time, 86° F.

The results of this study clearly indicate that very good chest radiographs can be produced well within the Federal guidelines of 30 mR for a 23 cm patient if the kVp and image receptor are carefully chosen (2). It was also found that it is not necessary to use a "rare earth" system or to use a high kVp in order to comply with the recommendations since excellent radiographs of the chest were produced using conventional screens ($CaWO_4$, $BaPbSO_4$, $BaSrSO_4$: Eu) at 80 kVp.



Dupont Par Screen
6 Plus Film
80 kVp

3M Alpha 12 Screen
XUD Film
80 kVp

Figure 1

RADIOGRAPHIC POSITIONING

The phantom has also proven useful in studying how changes in radiographic positions affect the visibility of certain lesions. A preliminary study using simulated pleural plaques of the costopleural margin indicates that very thin plaques (< 1 mm) can be demonstrated by various degrees of obliquity.

Initial results indicate that more shallow obliques (5° - 15°) are better for the visualization of plaques situated higher in the thoracic cage, and steeper obliques (35° - 45°) are better for the demonstration of thin plaques situated lower in the thorax. Work is continuing in this area in collaboration with Dr. Sargent at the University of Southern California. Other positions that can be studied are the lordotic view and stereo views.

OTHER USES

Since anthropomorphic chest phantoms have not been widely available, the extend of their applicability has not been fully explored. Potential areas of use include:

1. Producing radiographs for use in lung nodule detection experiments
2. Producing radiographs used in observer performance studies
3. Producing radiographs for studies designed to correlate objective criteria with radiologists' subjective impressions.
4. Phototimer set-up in chest units (e.g., measuring exit exposures)
5. An aid in teaching principles of radiographic technique to residents and technologists
6. Using a high contrast insert, the phantom can be used to compare radiographic techniques in the angiography suite.

OPTIMUM CHEST TECHNIQUE

The concept of an optimum chest technique has been discussed by several of the previous speakers at this symposium. It is generally realized that any considerations of an optimal chest technique must first define the radiodiagnostic task.

Another point that must be remembered is that a technique which will optimally image a certain pathologic entity in a standard patient must also have sufficient error margin (technique latitude) to adequately image a range of patient shapes and sizes (e.g., obese, muscular, frail, large-breasted).

SUMMARY

An anthropomorphic phantom with realistic vascular patterns and simulated pathology provides a clinically relevant method of evaluating and comparing images produced by different radiographic techniques. The results of these comparisons could be correlated with measurements of contrast, sharpness, latitude, speed and noise in order to bridge the gap between quantitative and qualitative assessments of image quality. However, the results of any method for evaluating radiographic imaging systems must be corroborated in the clinic before the ultimate usefulness of the method can be determined.

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DISCUSSION

MR. WOCHOS: The pneumoconiosis study used the Trout phantom; they found it very useful for quality assurance purposes and also in determining which facilities could participate in the study. With this phantom, they have been able to substantially decrease their rejection rate from what had been previously reported by Dale Trout and others.

DR. MOORE: I would like to address this question to any of the speakers. Has anyone been able to devise a method of creating any kind of realistic motions in the anthropomorphic phantoms?

DR. PEARCE: Like a lot of people, I used to think that the motion was important, but I think I have heard a theme this week that the importance of motion has decreased greatly because of the present short exposure times. However, the addition of motion to the phantom is not impossible; the idea is more than a twinkle in our eye. Nonetheless, it is difficult and I believe that if you do most of the basic imaging criteria with a suitable static phantom then you will have defined most of the parameters that are the problem. That should allow you to design a prospective study in the clinic with much more accuracy and then you can take physiologic motion into consideration. I think our efforts would be better spent on many other aspects of imaging rather than on trying to put a moving balloon inside the heart.

DR. MILNE: I wonder if I might add to that response. Dr. Pearce also did a simplistic but useful experiment to incorporate motion into our phantom. We are doing a study on real time videosubtraction for which we need motion in the phantom. Dr. Pearce's solution was quite ingenious; he slipped a tube through a third of the phantom with contrast medium flowing through it and then interrupted the flow with air bubbles to produce visual motion through the phantom. We, therefore, had superimposed lung structure and a normal thoracic wall so that there was normal scatter, while at the same time we had motion through the phantom at a controlled flow rate. That is one very simplistic way in which a phantom can achieve motion. Another is to place it on a device that makes it shake and oscillate with a known frequency.

DR. NICKOLOFF: I am clearly impressed with some of the work that is being done with phantoms, but one thing that disturbs me is that the standard phantom is of little use to evaluate the technique. I guess what bothers me is that the standard 23-cm patient with a fixed tissue composition has been simulated by the phantom, but some patients can be up to 30 or 35 cm in PA chest dimension. Low kVp techniques, for instance 70 kVp with no grid, may be very comparable to 120 kVp with grid when a 23-cm phantom is used. The techniques may not be comparable for much larger patient sizes, due to both attenuation and scatter. A second thing I noticed is that most of the exposure comparisons were done the way the Federal agencies have been doing them--free-air measurements with no backscatter. I find that free-air measurements made with probes result in reduced exposures. For example, you might get 15 mR with that kind of measurement, whereas with a real patient and TLD's on the skin, the measured exposure might be 22 to 25 mR, which is a 50 to 60 percent increase.

DR. CHRISTENSEN: Your second concern is much easier to address than the first. We did not actually measure the percentage of backscatter radiation, but the figures I gave you do include backscatter. The amount was taken from published tables and added to our measured exposures. In response to your first concern, I think the most important thing that we can do is to convey to people an enthusiasm for working on phantom projects. We do not have all the answers. I think this is just the first step. All I can say for certain is that the phantom we used produced these results. I would encourage you to design a better phantom and prove me wrong. I do not know if our results would be correct if we transferred them to a clinical situation, and any criticism you make of the phantom is almost certainly going to be valid, but I do not think that should stop us from trying to construct better phantoms.

DR PEARCE: It is doubtful if phantoms, at least in their present state of development, can answer all the questions raised about chest radiography. What I have suggested is that they allow you to define the problems. Last night you indicated to us that, because of government recommendations, you experienced some agony from the NCI in your institution and had all kinds of problems, and you seem to imply that you continue having practical problems with your current chest films. My point is that the phantom would at least help you sort out in what five directions you should go at once, instead of what 100 directions.

DR. MILNE: I do not agree with either of the previous responses even though Dr. Pearce is from my own lab. I do not think the phantom needs to be excused. The question by Dr. Nickoloff is a marvelous one and absolutely appropriate to this meeting; the phantom we have shown is ideally adapted to answering it. There is no reason at all why we should not distribute TLD's throughout the interior of the phantom thorax. It contains human blood; it contains human tissue; it contains air; therefore, the answers we derive would be completely valid. You are absolutely correct about what should be done. We should have TLD's distributed on the skin of the back, the front, and throughout the phantom. There is no reason why, with this phantom, we should not in this way obtain a superb spatial distribution of radiation within the lung. Although we do not need to excuse the phantom, you could equally well use Dr. Christensen's phantom or ours. This is a study that has not been done and it is a very important one to do. It also occurs to me that it would be interesting to study the marrow dose for an AP film versus the PA film. How long would it take you to do that? It would take about 1 week with this phantom, placing TLD's right in the center of the vertebral body and actually measuring dose in situ. So the phantom is absolutely adaptable for this study. Now, as to the point about the thickness of the patient, nothing whatsoever stops us from adding tissue equivalent material to the front and back of the phantom. What we need to do is accept that this work on dose distribution is a very good and valuable thing, then go ahead and do it!

DR. PEARCE: I will respond to the response to the response. I hope I did not imply that I was excusing the phantom. If you actually look at the appropriate literature, the TLD measurements on the phantoms have been done. Yes, we have what we call dry water. With these tissue equivalent slabs we can build the phantom up to any size and shape you like. We also use them instead of the buckets of water that you have been hearing about. They are much easier to put in a brief case and take with you; they are very consistent and highly calibrated; and they are not chlorinated, or whatever. I do not mean to excuse the phantom. I just think there are a lot of aspects to be considered.

DR. CAMERON: These are wonderful phantoms but there are only a couple of them in the whole country and in the world. They are not going to solve the general problem. I think there is a need for a less expensive phantom. I would suggest a "Rent-A-Phantom" service be established, since the typical hospital only needs a phantom when it is trying to set up a new unit or check out a new detection system.

DR. PEARCE: All I can do is agree with you. I think we are at the early stages of the value of phantoms in the whole diagnostic spectrum.

DR. AMTEY: I agree with Dr. Pearce. However, this phantom has the versatility and we all recognize its possibilities. While you are there, let me ask you, what is the delay in manufacturing and the price of something like this? We have been asking for it for almost a year and we have not gotten it. I admit this question is loaded, but could you tell me what the problem is?

DR. PEARCE: I can give you a loaded answer. I am not in the phantom manufacturing business. The company we are looking at had it for 2 years, and I am afraid the question should be addressed to them. It is not my immediate problem.

DR. AMTEY: In other words, are you saying it is a manufacturing problem? Since they worked with you on the development, I am wondering whether the delay is because of a desire to protect the technique or simply because of problems in manufacture.

DR. PEARCE: The technique is described in the April issue of Investigative Radiology (Pearce, J.G., E.N.C. Milne, and G.D. Gillan: Invest Radiol 14:181-188, 1979). Nothing is patented on this. You can make it yourself, so I assure you we are not holding back anything. I do not know if you have ever tried to develop anything and get it to market, but we have found that development is one thing and dealing with manufacturers is another game altogether.

A REVIEW OF TECHNOLOGIES AND METHODS FOR COMPUTER ANALYSIS OF IMAGES

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Computerized image processing has progressed rapidly in recent years, partly in response to overwhelming amounts of data. Meteorological satellites, the Landsat or Earth Resources Technology Satellite program, and planetary probe experiments have all produced huge quantities of image data in forms which allow and require computer handling. These projects have spurred development of other kinds of image analysis. In earth resources analysis, digitized aerial photographs can yield valuable information and airborne thermal line scanners are in use to monitor energy loss and power plant heated water discharges. Medicine, of course, has seen remarkable progress in computer aided tomography, ultrasound, and other techniques producing synthesized images in which machine processing plays a major role.

Technological advances in several areas have had major impacts on image processing. Notable are advances in television communications, including video recorders and "instant replay" video disks, and in computer memory devices.

A typical present-day image processing system will be configured something like that of Figure 1. Its major components are a television monitor for image display, an image store area, a computer to process images and to control the system, and some means such as keyboards and joy sticks to allow operator interaction. The monitor is used to form a raster

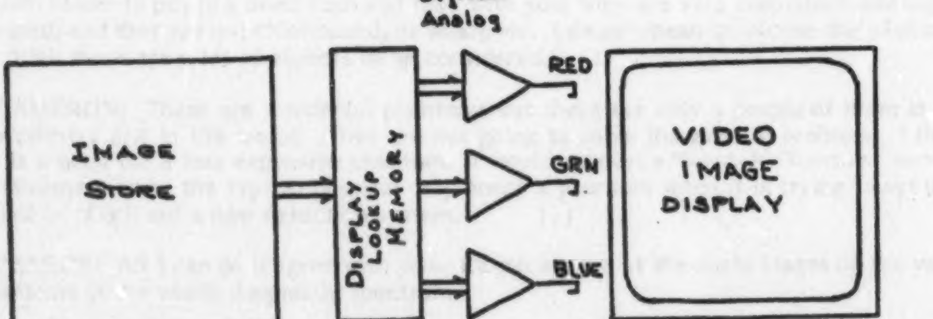


Figure 1. Major components of an image processing system.

scan display which is refreshed periodically (usually 30 times per second). Each raster is created by cycling digitally stored numbers past digital-to-analog (D/A) converters, producing time-varying analog voltages which drive the electron guns in the monitor in synchronism with the refresh characteristics of the display. The image data themselves reside in the image store with a row for each raster and an appropriate number of samples per row. Since these values are associated with specific points of the display, they are called picture elements, or "pixels." Common configurations use 256 x 256 or 512 x 512 pixel displays and storage areas, and often have room for several independent frames.

Generally a fixed amount of information is stored for each pixel. This may be a single value from perhaps 0 to 63 or 0 to 255, stored in 6 or 8 binary digits, or bits, or it may be more complex. Storage for each of several multispectral channels may be desirable, or one or more bits per pixel might be dedicated to single value graphics overlays to outline an object, superimpose a map, or create a cross hair or rectangle controlled by a joy stick.

A small memory can be provided as a lookup table between the image store and the D/A converters to establish the color, brightness, and contrast that are to appear for a given image value. The characteristics of this table can be altered by passing very small amounts of information from a controlling computer, so that displayed image colors or brightnesses can be changed easily and quickly. This allows very inexpensive image enhancement because the data in the image store need not be altered.

Several technologies can be used to provide the image store. Video disks have been used, allowing storage and rapid recall of up to several hundred frames. Digital-to-analog conversion at very high speeds allows the data to enter the system in digital form. This approach is useful for multiple image studies or animation, but radiometric quality is poor. Digital disks can be used, storing several frames per disk; the chief drawback of this system is the lack of processing flexibility. Electronic memories using solid state integrated circuits are currently the most popular storage technology; their costs, though dropping, are high enough that multiple frame stores are uncommon.

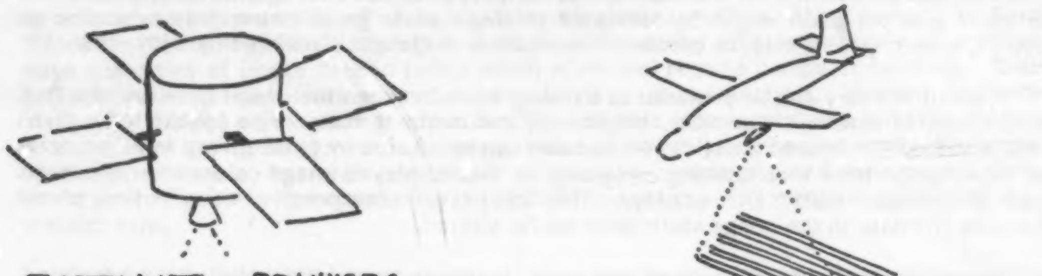
Digital image analysis is only possible if the image data are in numeric form. Satellite sensors provide data in this form, while most other data sources demand a preliminary image digitization step (Fig. 2). Virtually all analog-to-digital conversion processes require periodic sampling of a time-varying signal. This can be created from an image with television techniques, but usually at the expense of radiometric quality. Another technique uses a scanning densitometer, in which transmitted or reflected light is sensed as an image is moved with respect to a light source. These instruments commonly possess very good geometric and radiometric fidelity, but are often rather slow.

Many kinds of image processing involve transformations of the data themselves, requiring extensive use of a computer. In most hardware configurations, the image store is not directly available to the computer. It must work with an external copy of the image (probably on tape or disk), process it piecemeal, and eventually write the transformed data to the image store. This process is slow and expensive, but progress seems likely. Some of the newest systems do have randomly addressable image stores. Also, "array processors," which perform identical operations simultaneously on all cells of a large array, may soon be commercially available.

Many kinds of computerized processing are possible. Enhancements, or "contrast stretches," are often fairly simple. New brightness values can replace originals according to some rule, such as equalizing the overall distribution of grey levels. Edge enhancement can be achieved by seeking sudden changes and synthetically increasing image brightness at those points. Edge-enhanced satellite images are proving useful to petroleum geologists for lineament detection. Image resampling is often done to create new images which are geometrically corrected, perhaps to overlay satellite data and maps. Resampling should be done with care and thought, for it is a potential source of considerable noise and error. Multiple scenes, if geometrically true, can be overlaid to seek similarities or differences.

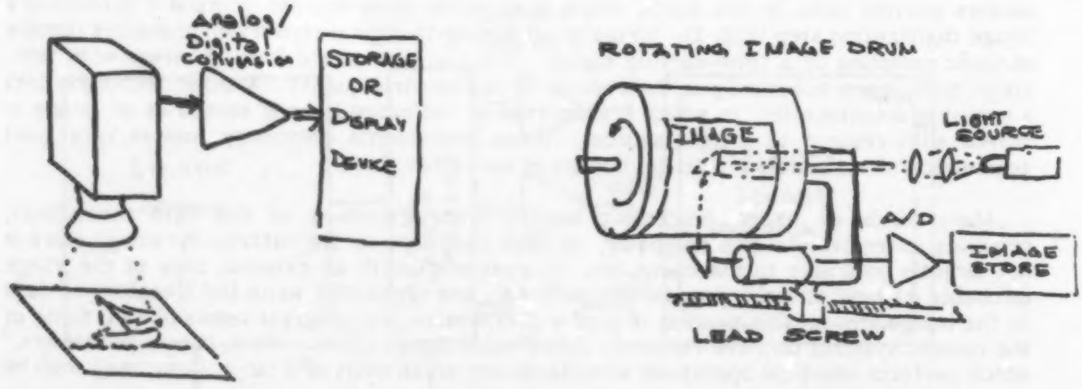
Mathematical transformations on multispectral images can produce "synthetic channels" which maximally separate specified features. Classification algorithms are frequently used by remote sensors to assign each pixel of an image to one of a finite number of classes, usually by selecting a statistically "best" class.

Image processing technology is moving out of its infancy. The equipment and procedures are bound to become more common and more useful in virtually all areas dealing with images.



SATELLITE SCANNERS
(LANDSAT; GOES; SMS)
ROTATING SCANNERS OR
OSCILLATING MIRRORS--
ON-BOARD DIGITIZATION
BEFORE TRANSMISSION

AIRBORNE SCANNERS
(THERMAL OR MULTISPECTRAL)
RECORD ON MAGNETIC TAPE
WITH ON-BOARD OR GROUND-BASED
A/D CONVERSION


Analog/
Digital
Conversion
STORAGE
OR
DISPLAY
DEVICE
ROTATING IMAGE DRUM
IMAGE
LIGHT
SOURCE
A/D
IMAGE
STORE
LEAD SCREW

TELEVISION CAMERA
IMAGE CONVERTERS

ROTATING DRUM
SCANNING DENSITOMETER

Figure 2. Image digitization methods.

COMPUTERIZED SYSTEM FOR INTERPRETATION OF RADIOGRAPHS FOR PNEUMOCONIOSIS

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INTRODUCTION

Our review of both the current literature and our present image analysis system suggested that pre-processing of radiographs is an essential part of computerized interpretation. Previous work by Hall et al. did not incorporate this pre-processing, which probably accounts for their difficulty in repeating the results. In addition, the recent results of several other investigators (University of Pittsburgh, Purdue University, University of Missouri, and Queens University of Canada) should be incorporated into our system before final testing. A faster computer will be used to reduce the time required to process a radiograph from 60-70 minutes to less than 6 minutes. An automated radiograph handling setup will allow the system to operate continuously on a 24-hour basis.

The significant contribution to computerized radiograph interpretation that this study will make over previous studies is in the large number of films which will be analyzed. Using algorithms developed by us and by other investigators, we plan to analyze more than 2,000 films, 1,000 of which have been interpreted by at least three B-readers.

HISTORY OF SYSTEM FOR COMPUTERIZED INTERPRETATION OF CHEST X-RAYS

NIOSH's support for the computerized interpretation of chest x rays began with NIOSH Contract No. HSM 99-72-30 and continued with Contract No. HSM 99-73-92, both with the University of Southern California. Previously published results (1,2) of the principal investigators, Ernest L. Hall and Richard P. Kruger, provided a reasonable basis for NIOSH to fund the contract in order to extend the investigators' work to coal workers' pneumoconiosis.

Results of the contracts, published in various journals (3-5), demonstrated that it is possible to automatically interpret chest x rays for pneumoconiosis with "classification results comparable to experienced radiologists." Results for a normal/abnormal classification, using a training sample of 130 films, gave 92 percent correct classification for the 130 training films and 74 percent correct classification for a set of 338 films not previously seen by the computer. However, the contractor was later unable to repeat these results; the system could not "pass" the B-reader examination. All films in the original study were scanned or processed within a short period, and the contractor felt that the system was not able to repeat the results due to the instability of the system. Some attempts were made by the contractor to correct these problems, but they were not completed before the contract expired. In October 1976, the system was delivered to Morgantown for further testing and evaluation.

"IN-HOUSE" PHASE OF THE PROJECT

The first step was to conduct a complete evaluation of the stability of the system. The results of our stability tests indicated the need to compensate for variations in the analog hardware characteristics. In a recent Ph.D. dissertation (6) and later publication (7), Richard Conners, University of Missouri, stressed the importance of normalizing image contrast of radiographic images through the use of a linear distribution function or histogram equalization. Normalization is necessary because of the limitation of the dynamic range of the image scanner and significant variation in density values from one radiograph to another. No normalization of the radiographic images was previously performed, and the omission of this pre-processing of the radiographic image could very well explain the contractor's inability to repeat the results and "pass" the B-reader examination.

Figures 1 and 2 show the gray level histograms for two normal films with acceptable film quality. Note particularly the significant differences between the two gray level histograms. If only the histograms for gray level within the lung field of both films are considered (Fig. 3), the difference is apparent. Histogram equalization, or normalization of the gray level histograms, would force all of the x-ray films to have the gray level histogram shown in Figure 4. This means that although the original gray level histograms are different, with histogram equalization the gray level histograms of the lung fields would be constrained to be equal for all films before texture measurements were derived.

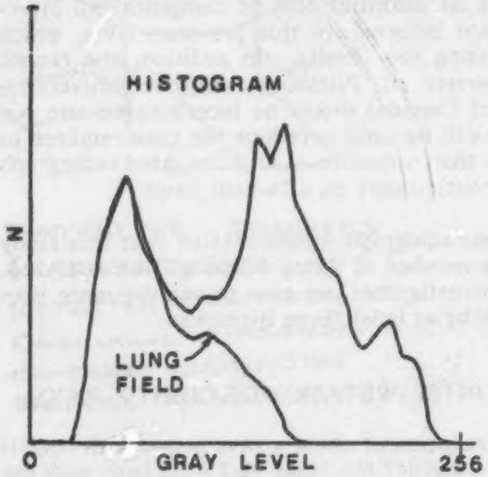

Figure 1. Gray level histogram of normal x-ray film with good film quality (total film and lung fields only).

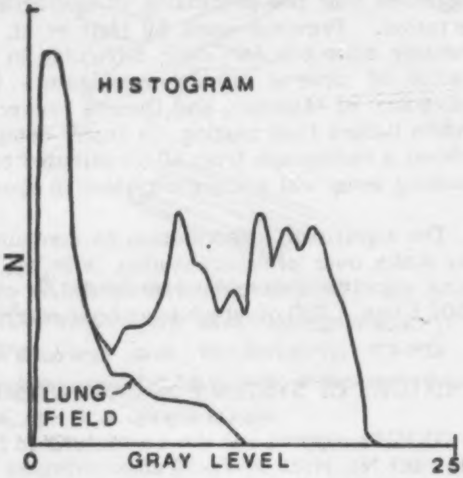

Figure 2. Gray level histogram of normal x-ray film with acceptable film quality but slightly overexposed.

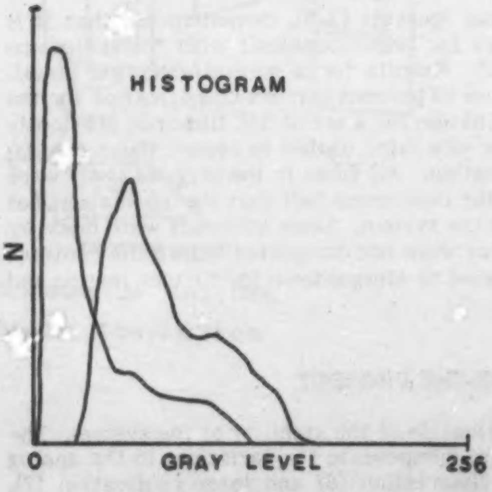

Figure 3. Gray level histograms of lung fields of x-ray films in Figures 1 and 2.

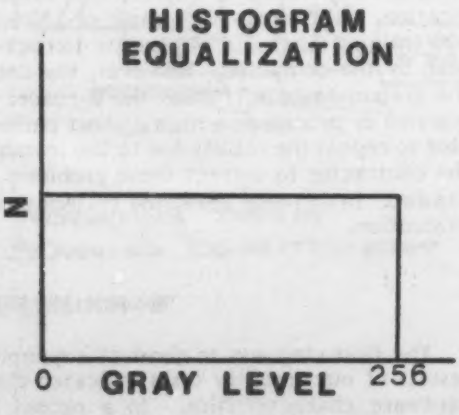

Figure 4. Gray level histogram of x-ray film after histogram equalization.

REVIEW OF RELATED STUDIES

1. **The original results of Hall and Kruger were probably not repeatable because they did not compensate for variations in the scanner or normalize the radiographic image (3-5). Further software modifications and additional software could solve these problems and allow us to repeat or improve on their results. Figure 5 shows a typical "confusion" matrix which is used to describe the accuracy of the classification results. Figure 6 shows the results of the Hall and Kruger study for a normal/abnormal classification, using an average of the five readers, the training sample of 130 films (training results), and a set of 338 films not previously seen by the computer (test results).**

INDIVIDUAL READER

		N	A
TRUE CLASS	N	TRUE NEGATIVE	FALSE POSITIVE
	A	FALSE NEGATIVE	TRUE POSITIVE

Figure 5. Method of describing the accuracy of classification results - "confusion matrix."

AVERAGED READER RESULTS

		N	A	
CONSENSUS	N	96.3%	3.7%	91.6%
	A	10.8%	89.2%	

TRAINING RESULTS

		N	A	
CONSENSUS	N	86%	14%	92%
	A	4%	96%	

TEST RESULTS

		N	A	
CONSENSUS	N	48%	52%	74%
	A	16%	84%	

Figure 6. Results of Hall and Kruger study for a normal-abnormal classification.

2. Recently, C.C. Li et al. demonstrated a successful algorithm for detecting individual small opacities appearing during the early stages of pneumoconiosis (8). The study, funded by a grant from the National Science Foundation and the State of Pennsylvania, was aimed at early detection of pneumoconiosis by counting and sizing small opacities which appear on the chest x ray.

3. Jagoe and Paton, using a mean "diversity" measurement, were able to correctly classify 80 percent of 36 chest x rays (9 in each pneumoconiosis category) (9). This algorithm could be added to our system.

4. Ballard and Sklansky described an algorithm which successfully locates tumors in chest radiographs (10). Although the algorithm uses a considerable amount of computer time (5 hours on an HP 2114B), the algorithms they used could be added to our system, if necessary, with computer hardware improvements.

5. Kruger et al. reported even better pneumoconiosis classification results when they looked only at the regions between the ribs, using a manual method of locating these regions (3). At the time of their study, there were no published algorithms for automatically detecting ribs. However, several successful rib detection algorithms have recently been developed (11,13). The addition of rib detection algorithms and rib removal could significantly improve our system accuracy, but would require significantly more computer time.

6. Toriwaki et al., of Japan, have developed a computerized system which detects abnormal chest x rays (14-16). A much more sophisticated system than ours, it eliminates the ribs and then screens, not only for what they refer to as "shadows" in the lung, but also for abnormalities in such items as heart size. Published results suggest that the system is used in a mass screening environment.

Since completion of the University of Southern California contracts, other investigators (6-11, 13-16) mentioned above have published results which provide some support for the findings of Hall and Kruger. The only problem associated with the algorithms used in these studies is that they require a considerable amount of computer time; therefore, for the algorithms to be practical, a fast computer is needed. Because most of these investigators used rather slow computers, their studies usually have processed only a relatively small number of films. A larger study involving at least 1,500 films is needed to prove that these algorithms will work on the types of films being collected in surveillance programs. It is also clear that any study using a large number of films, with a corresponding large variation in density values, must normalize the radiographic image (pre-process) before any measurements can be taken.

USE OF COMPUTERIZED INTERPRETATION SYSTEM

Once the system has been validated, it will be used by NIOSH in both its research programs and its medical surveillance programs. Currently, the coal miner medical surveillance program receives approximately 130,000 chest x rays per round of examinations. Each film is interpreted at least twice--once by an A-reader or B-reader and again by a B-reader.

The system for computerized interpretation of chest x rays could be used in the medical surveillance program in the following manner:

1. The film would be interpreted immediately by an A-reader or B-reader and, if it is judged abnormal, it would go to a second B-reader for interpretation.

2. If the first reader interprets the film as normal, it would go to the computer system for interpretation.

3. If the computer system interprets the film as abnormal, it would go to a B-reader for interpretation.

4. If the computer interprets the film as normal, agreeing with the first reader, no further interpretation would be needed.

Because approximately 90 percent of the chest x rays are normal, potentially we could eliminate 110,000 second interpretations per round of examinations. At $2 per interpretation, this would amount to an estimated potential savings of $220,000 per round of examinations, whereas the computerized system cost approximately $120,000. Eliminating B-reader interpretation of normal films would also utilize the B-readers more efficiently by allowing them to interpret only suspected abnormal films. The computer would simply serve as a check on the first interpretation, which should be an acceptable function if the computer system has successfully passed the B-reader examination.

In addition to its use in the surveillance program, the system will be used in NIOSH research activities. It could serve as a consistent, unbiased interpreter of chest x rays, giving both zonal and overall readings. The system could also be used in research into other occupational respiratory diseases.

The results of this research project could also serve as a model for the use of computers to streamline the interpretation of medical data. One criticism of required medical programs for occupational surveillance is the burden they place on our nation's health care delivery system. The funding and completion of this study demonstrate NIOSH's awareness of this problem and its willingness to resolve it.

STEPS IN PROCESSING A CHEST RADIOGRAPH

Listed below is a detailed description of the steps in scanning and processing a chest x ray and the corresponding processing times on a Digital Equipment Corporation PDP 11/40 computer. Figure 7 gives a flow chart of these steps:

1. Scanning of the chest x ray: 3 to 5 minutes.

   The time required to scan an x ray is based on observations of the current system. The scanning time cannot be improved because of analog hardware limitations. However, the system could be made more efficient by doing processing and scanning simultaneously.

2. Pre-processing of the image (using an algorithm described by Conners (6,7) or histogram equalization): 10 minutes.

   Pre-processing of the image is necessary because of drift and aging of the scanner and variability in the technique used to produce the chest radiographs. The time required is based on disk input/output time and experience with similar algorithms implemented at the Appalachian Laboratory for Occupational Safety and Health.

3. Rib detection: 22 minutes (ribs may be removed later, if necessary).

   It may be necessary to compensate for the detrimental effects of the ribs on the texture algorithms. Some investigators conducting research in image analysis of chest x rays remove the ribs as part of pre-processing (8, 12, 14-17). The estimate of time required is based on times quoted by Wechsler and Fu of 18 minutes using a PDP 11/45, which is at least 20 percent faster than the PDP 11/40 (17).

4. Texture measurements: 20 minutes.

Most of the current processing time is associated with the texture measurements. Because of the time required to make these texture measurements, previous investigators have written the software in Assembly Language. We have rewritten the texture measurement algorithms in FORTRAN. The FORTRAN program required 59 minutes to execute on the PDP 11/40; this time could be shortened by computing the texture measurements only in the lung fields.

5. Other measurements: 8 minutes.

Several additional measurements would be useful in classifying an x ray. The time required to make these measurements is difficult to estimate.

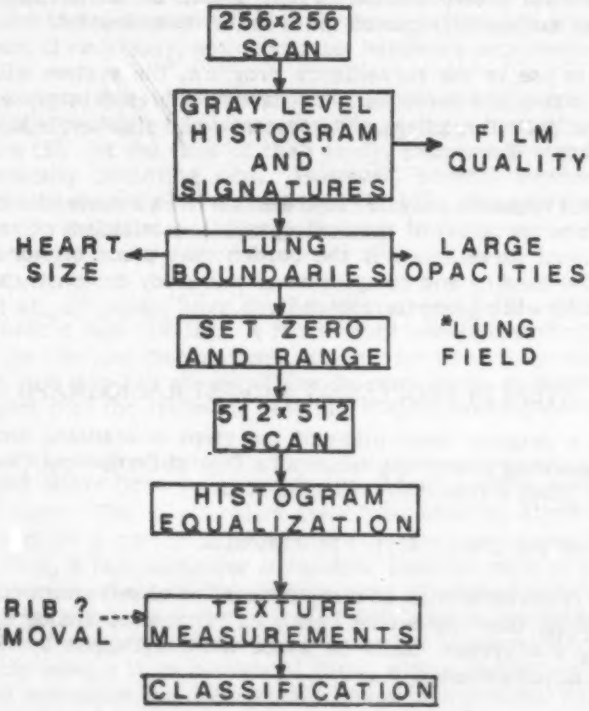
256x256
SCAN
GRAY LEVEL
HISTOGRAM
AND
SIGNATURES
FILM
QUALITY
HEART
SIZE
LUNG
BOUNDARIES
LARGE
OPACITIES
SET ZERO
AND RANGE
*LUNG
FIELD
512x512
SCAN
HISTOGRAM
EQUALIZATION
RIB ?
REMOVAL
TEXTURE
MEASUREMENTS
CLASSIFICATION

Figure 7. Flow chart of steps used to process a radiograph by NIOSH image analysis system.

The estimated total time to process a chest x ray is thus 64 minutes on the PDP 11/40. Consequently, the present system would require a minimum of approximately 2,133 hours just to process the minimum of 2,000 chest x rays needed to train and test the system. In addition, the algorithms described 2 and 3 above must be further developed on a large number of chest x rays interactively with the image display system. The long processing times make this interactive program development extremely time-consuming. The use of a faster computer and an automated film handling system will allow us to reduce the processing time to a reasonable value of less than 6 minutes. For example, the FORTRAN texture measuring program described in Step 4 above required 59 minutes to execute on the PDP 11/40; however, it required only 2 minutes and 31 seconds to execute on a Digital Equipment Corporation VAX 11/780.

FUTURE POSSIBILITIES

Further work could be directed toward development of an unsupervised classifier for the machine which, after initial training, would correct itself as it classified films. The machine would adjust its parameters so that it could, in effect, train on a much larger set of films. This could result in a self-learning machine and would allow us to partially overcome the lack of a well-defined training set. Unless the machine has some self-learning capabilities, its accuracy will always be limited by our ability to provide a large, accurate set of training films. The development of an unsupervised classifier has been described in several texts (17-20).

QUESTION

MR. SHAVER: Can you tell us what you meant by "normal?"

DR. HANKINSON: Normality was defined as no pneumoconiosis present (category 0/1 or less). If there was any pneumoconiosis present (category 1/0 or greater) in any of the 6 zones, the film was classified as abnormal (4).

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PERFORMANCE OF ELECTRON RADIOGRAPHY IN CHEST RADIOGRAPHY

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INTRODUCTION

A prototype automatic chest unit, using an electron radiographic detector system was used at the University of Southern California for about 1 year around 1975 (1). This note is a brief summary of our experience with this unit.

TECHNICAL DETAILS

The electron radiographic detector used in the chest unit employed xenon at about 8 atm. absolute pressure. The electrode spacing was about 10 mm and the electrostatic field in the detector was focussed on the x-ray tube focus spot electrostatically (2). For more details on electron radiographic detectors the interested reader should consult the work reviewed by Seelentag (3) or by Muntz and Jacobson (4). The electron radiographic chest system was designed and constructed by Xonics, Inc. The characteristics of the system were such that it required about 0.5 mR to give unit optical film density. The equipment possessed the capability of significant analog image processing because of the nature of the electrostatic image. Varying degrees of image discharge, and consequently varying "edge effect," could be obtained, as well as scatter suppression using a bias potential during the processing step. The technical details of these possibilities are given by Muntz et al. (5,6).

PERFORMANCE

The particular system that was used produced quite acceptable radiographs (when it worked) with about half the exposure required for a conventional par speed film/screen combination. It was found that a certain amount of "edge effect" (corresponding to about 60 percent image discharge), was desirable, but excessive amounts were undesirable. This was because the emphasis on high frequencies caused excessive normal tissue clutter that tended to obscure larger area targets of interest (e.g., centimeter size and larger nodules). The flexible contrast capabilities made possible by the electrostatic processor were useful and were used to eliminate scatter simply by applying a bias potential in the image processing step (6).

Because of the characteristics of the detector, very high kilovoltage chest radiography (greater than 120 kVp) was not very satisfactory. The resolution of the electron radiographic detector depends on x-ray photon energy, with the resolution degrading noticeably at higher beam energies. An example of a radiograph obtained with the electron radiographic chest unit can be seen in Figure 9 of the article by Muntz et al. (6).

SUMMARY

Electron radiography does provide satisfactory chest radiographs. The analog image processing capability of the image developer was useful and is an attractive feature of electrostatic chest imaging.

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POTENTIAL OF COMPUTED RADIOGRAPHY FOR THE CHEST

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ABSTRACT

The General Electric prototype computed radiography (CR) system uses a narrow fan beam, xenon gas detection with wide dynamic range, and digital image processing. This results in greatly reduced scatter, increased exposure latitude, and capability of quantification and image manipulation to extract the information contained in the radiographic signal. The spatial and contrast resolution of the CR system (1,2) are discussed and compared to screen/film systems to show the potential (3) of CR in chest radiography.

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Note: This abstract is presented in lieu of the manuscript, which was not available at the time of publication. Further information may be obtained from the references listed above.

ILLUSIONS AND PERCEPTIONS IN RADIOGRAPHIC IMAGES

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Recently there has been considerable interest in applying computer processing techniques to radiographic images. The use of numerical manipulation algorithms introduces the potential for image phenomena which are not experienced in ordinary geometric imaging systems. This presentation discusses two aspects of illusion and perception which can occur; the first is generated by procedures which introduce a correlated or anticorrelated noise texture into the displayed image, and the second is a contrast illusion effect generated by the introduction of edge-enhancement steps in image processing. In our laboratory we have been most extensively concerned with computed tomography images; consequently, this presentation is primarily oriented to CT results. The general ideas, however, apply to any imaging system.

An example of the noise-texture phenomenon is shown in Figure 1. The CT images shown are from an anthropomorphic CT body slice phantom for which the spatial properties of the observed structures can be externally measured with great precision. The images on the left are the normal size and a factor of two expansion of a scan reconstructed with the standard manufacturer supplied algorithm. The images on the right are from a scan taken at about one-fourth the exposure and reconstructed with a modified algorithm of our invention. The pixel variance in the two cases is about the same, but the image on the left has much finer and more detailed noise-texture. The resolution and spatial acuity of the two images, however, is essentially identical; only the random noise spatial properties differ between the two. We have found the overwhelming subjective impression of viewers is that the image on the left is the "superior" image. It appears crisper, more definite, and seems to give higher edge acuity. Objective experiments, nevertheless, indicate that the "superiority" of the left-side image is an illusion. An observer can be asked, for example, to draw in the contour of the right margin of the liver. There is great subjective confidence in performing this task on the left-image blow-up; however, there is some distaste for the attempt on the right-side image, which does not exhibit the same sharp definition. In this case, however, the actual contour can be measured and the observer-estimated contours compared with the true values. The result is that the mean squared spatial errors between observer estimates and true values are larger for the left-side, "high-quality," than for the right-side, "lower-quality," image.

A hint at the origin of this effect is gained by studying the noise power spectrum of the image. This is shown in Figure 2. The standard algorithm produces the noise power plots shown. For the system presented, the MTF squared (i.e., signal power) falls to 10 percent of its initial value at about 0.3 mm^{-1}, whereas the plots show that the noise power is at its maximum value at this frequency. That is, there is a pronounced detail of noise texture which is much finer than the characteristic dimensions over which the system can resolve real structures. Figure 3 shows the image-space presentation of this same consideration by means of the noise autoconvolution function, the Fourier transform of the noise power spectrum. For the high-definition image on the left side of Figure 1, the solid curve shows that the noise is large and anticorrelated. The smoothed image, the right-side image of Figure 1 (the dotted curve) shows a smaller peak noise, more smoothed out, and less textured noise detail. For equal exposures on scans processed by the two algorithms, the two-dimensional spatial integral of noise would be approximately equal, but the regular processing has high anti-correlation properties. Our interpretation is that the presence of this kind of noise texture signals the eye/brain system that the image is characterized by much higher detail and acuity than is actually the case. An observer, therefore, suffers from an illusion defect in perception; in these systems, very fine noise texture does not imply that the definition and acuity of the image presentation of real subject structures is truly superior.

The second fascinating illusion has to do with edge-enhancement algorithms. Figure 4 shows a rough drawing to guide the discussion. The sketch on the upper left indicates the smooth, well-matched, system response to a sharp edge. The sketch on the upper right shows the negative of the Laplacian (two-dimensional second derivative) of that edge response. The sketch on the lower left indicates the result when a certain amount of image Laplacian is subtracted from the original image. The edge-response is enhanced. The sketch on the lower right indicates the MTF modifications which occur. Figure 5 shows the line spread functions of the high-definition and low-definition images generated by our scanner. We note that the high-definition response seems to be just the same as the low-definition response, but with some second derivative edge enhancement added. The result is an appearance of higher edge acuity when the image is ideal and noise-free, but the application of Laplacian processing to noise gives unpredictable results. An illustration of the sort of thing that can occur when edge derivatives are added to images is shown in Figure 6. The contrast disc that appears to be in the center of the image field is not really there. What is in the image, at the apparent edge of the disc, is just the edge Laplacian extending over a distance of a fraction of a millimeter. The second image presented is, in fact, just a copy of the first to which a black ring overlay is added to mask the "edge" variations. We see that the apparent image is actually no more than a small bipolar edge Laplacian located very near the black ring overlay. Yet, in viewing the entire image it seems certain that the interior of the circle is of significantly different brightness than the background. It seems obvious that such phenomena could greatly confuse interpretations of radiographic images in actual practice.

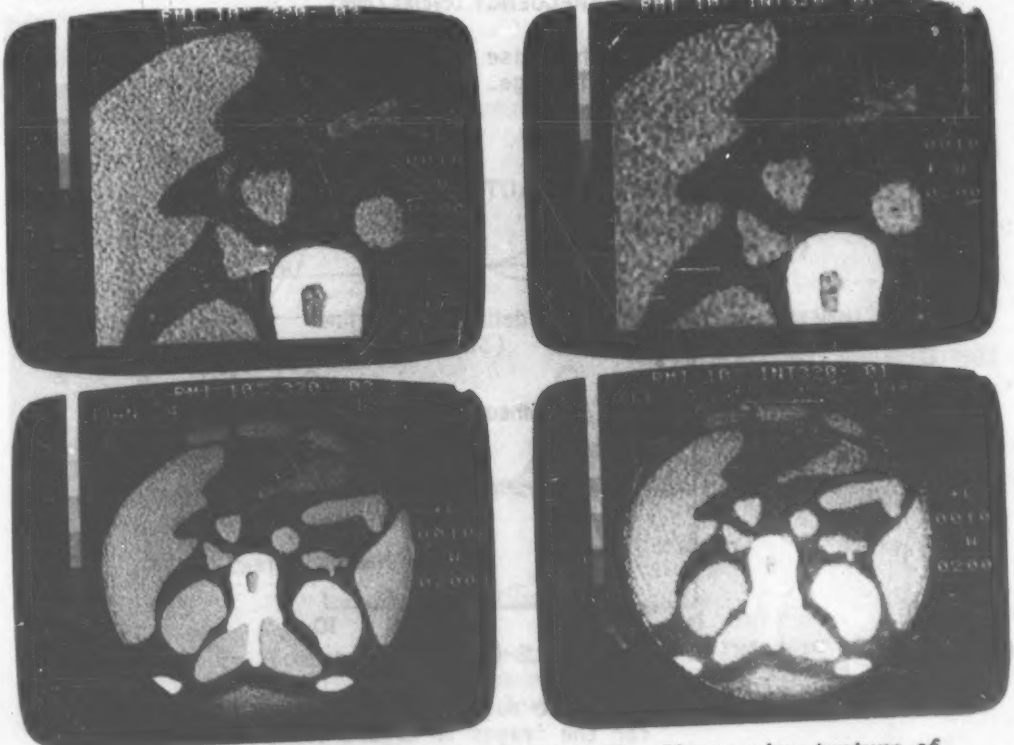


Figure 1. The effect of noise-texture. The finer noise-texture of the left-side images give the subjective impression of greater detail and sharpness. Objectively, however, the right-side images have equal acuity and less noise.

best copy available

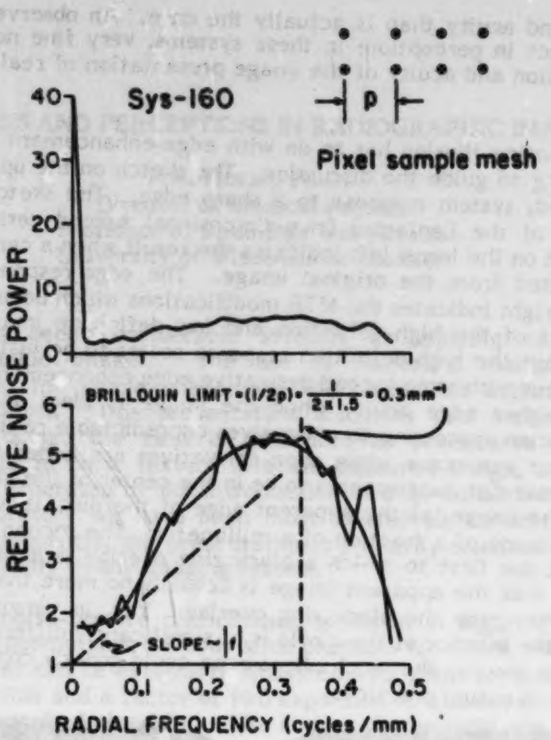


Figure 2. The noise power spectrum of a CT image.

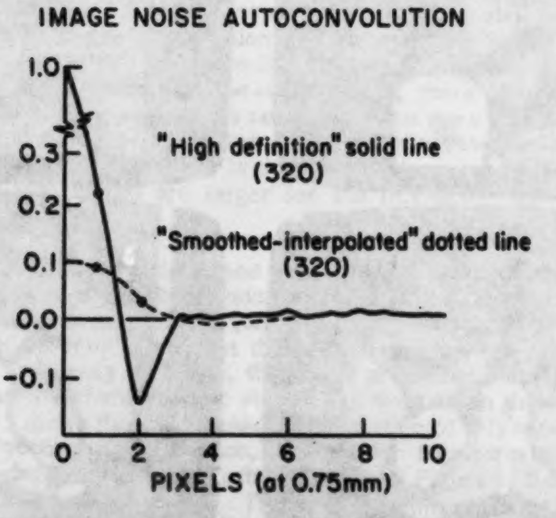


Figure 3. The noise autocorrelation behavior for the images in Figure 1. The solid line corresponds to the left-side images of Figure 1, the dotted to the right-side images.

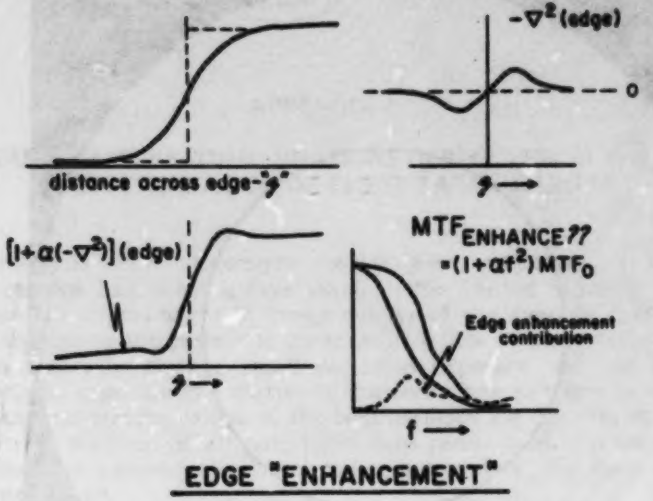
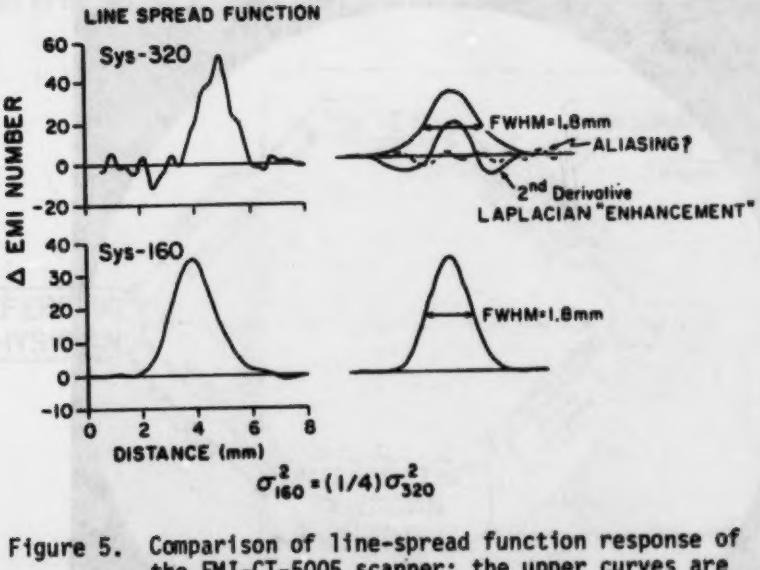

distance across edge-"ρ"
-∇² (edge)
0
ρ ⟶
[1+α(-∇²)] (edge)
ρ ⟶
MTF ENHANCE??
=(1+αf²) MTF0
Edge enhancement contribution
f ⟶
EDGE "ENHANCEMENT"

Figure 4. Illustration of the process of edge-enhancement of an image by computer manipulation of image data.



$\sigma^2_{160} = (1/4)\sigma^2_{320}$

Figure 5. Comparison of line-spread function response of the EMI-CT-5005 scanner; the upper curves are SYS-320 mode, the lower are SYS-160.

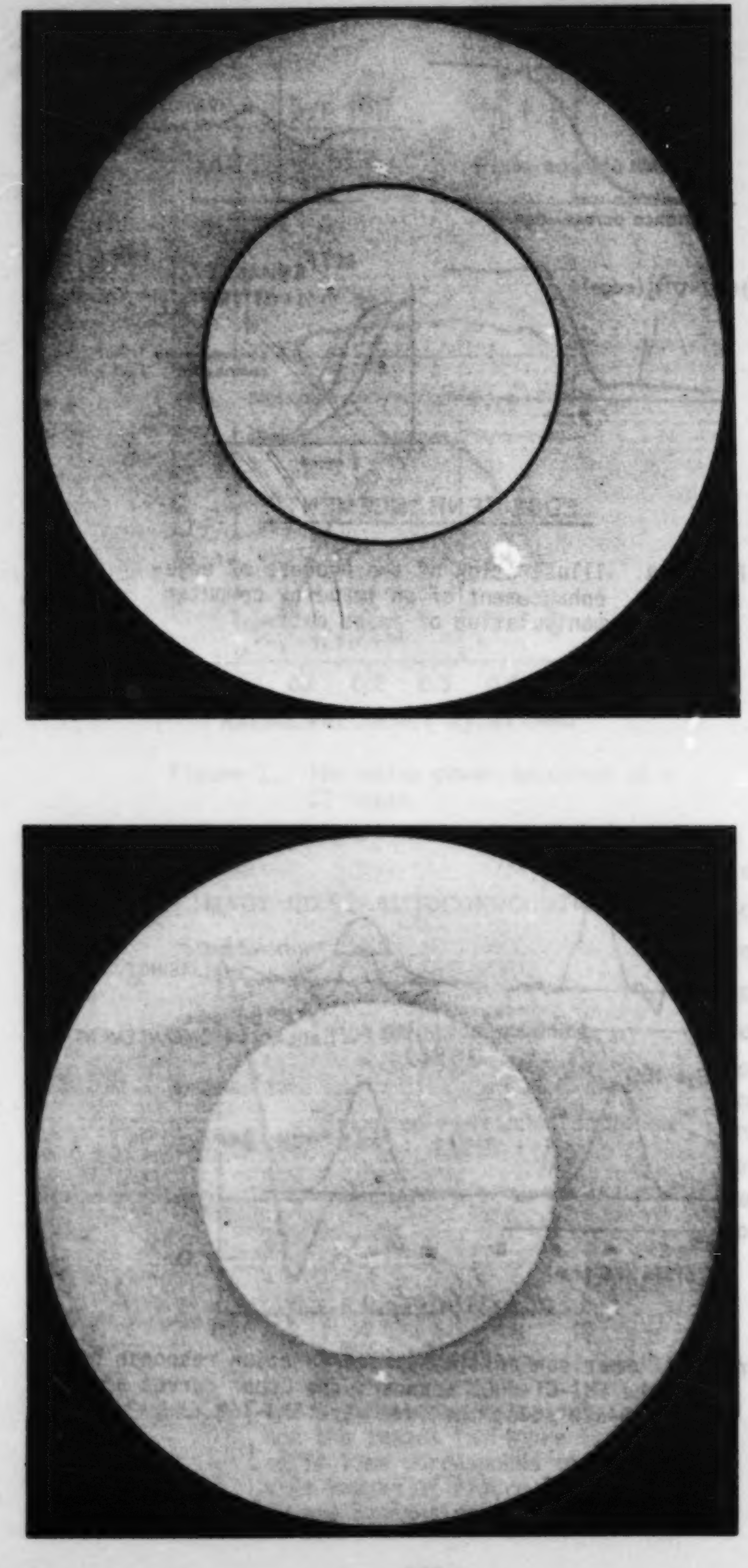


Figure 6. Contrast illusion generated solely by edge effects.

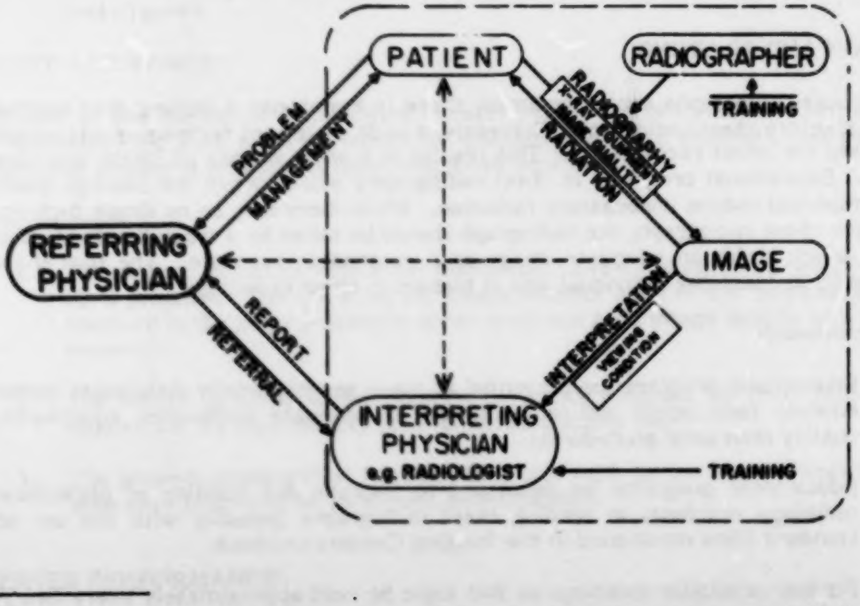
APPENDIX A

CONCLUSIONS AND RECOMMENDATIONS DERIVED FROM THE SYMPOSIUM ON OPTIMIZATION OF CHEST RADIOGRAPHY*

Chest radiography is the most common medical x-ray procedure. It is estimated that seventy million persons had chest x rays taken in the United States in 1978. There is considerable room for improvement in image quality of the average chest radiograph and the amount of exposure to the patient is unnecessarily high in many facilities. Although a chest radiograph typically gives a small radiation exposure and has little risk to an individual patient, the exposure is a matter of concern since so many people receive chest radiographs. Chest radiography is one of the best methods for discovering pathology in the chest. Since a large fraction of all hospitalizations result from disease of the lungs and heart, the importance of a good quality chest radiograph in detecting disease in these organs cannot be overemphasized.

The chest radiographic process involves a patient, a radiographer, a radiologist, and generally a referring physician. The interactions of these individuals are shown schematically below.

These recommendations do not cover all aspects of chest radiography. Our recommendations will concentrate on the aspects inside the dashed line. We considered the need for medical imaging laboratories, educational programs, reduction of patient exposure, quality assurance programs and techniques to improve observer performance in chest radiography. Some of these topics overlap and there are important topics which are not covered by these recommendations. These areas are ignored not because they are unimportant but because they were felt to be outside the general purpose of the symposium.


PATIENT
RADIOGRAPHER
TRAINING
PROBLEM
MANAGEMENT
RADIOGRAPHY
X-RAY EQUIPMENT
IMAGE QUALITY
RADIATION
REFERRING PHYSICIAN
IMAGE
REPORT
REFERRAL
INTERPRETATION
VIEWING CONDITION
INTERPRETING PHYSICIAN e.g. RADIOLOGIST
TRAINING

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* Developed by the planning/advisory committee following open workshop/discussion sessions.

IMAGING CENTERS

It was apparent from the papers presented during the symposium and from discussions during the workshops that further research is required in many areas in chest radiography, such as: technique, equipment, observer performance, and development of radiographic chest phantoms.

We recommend:

1. Funding be provided for a number of research centers (at least 4) specializing in radiographic imaging. Such centers should combine expertise from, and provide for communication among, radiologists, physicists, engineers, psychophysicists, computer scientists, and other appropriate individuals;

2. Further development of methodology for observer testing, such as ROC analysis;

3. Development of a standard set of chest films demonstrating a variety of diseases (e.g., tumors, heart failure, emphysema, pneumoconiosis) to be used in training and for observer performance testing;

4. Development of an inexpensive radiographic chest phantom that could be made widely available for evaluating chest radiographic equipment including proper operation of phototimed equipment; and

5. Development of a sophisticated, versatile, anthropomorphic chest phantom to be used to relate physical measurements to radiologists' interpretations.

The inexpensive chest phantom should simulate the contrast, resolution, scattering properties, and range of densities encountered in the human thorax.

The new information developed at these centers should be made available through scientific meetings and symposia. The imaging laboratories should also act as consultation centers with support and referrals to radiographic installations in these regions.

EDUCATION AND TRAINING

Since chest radiographs are so common, there is sometimes a feeling that anyone can take a satisfactory chest radiograph. Currently, a wide variety of techniques and equipment is being used for chest radiography. This results in a wide variety of image qualities and exposures. Educational programs in chest radiography will improve the average quality of the radiograph and reduce unnecessary radiation. While there may be no single technique to optimize the chest radiograph, the radiograph should be taken by a trained individual with a technique which yields an adequate image at a reasonable exposure. The film should be interpreted by at least one individual who is trained in chest radiology.

We recommend:

1. Educational programs be presented at many geographically distributed centers to educate radiologists and radiographers on adequate techniques, equipment, and quality assurance procedures;

2. Educational programs be developed to improve the training of physicians and radiology residents in reading chest radiographs (possibly with the use of the standard films mentioned in the Imaging Centers section);

3. Further scientific meetings on this topic be held approximately every two years; and

4. Educational materials be prepared, to inform the public on chest radiography. They should include information on radiation exposure, radiation risk, and the medical usefulness of a chest radiograph.

PATIENT EXPOSURE

The exposure to an individual patient from a chest radiograph is quite small and has only a very small probability of causing damage. However, the number of chest radiographs taken each year in the United States is large.

We recommend:

1. The criteria for requesting the chest radiograph be studied scientifically;
2. The optimum number of views or projections to be taken be studied and recommendations be made as to the appropriate views for various classes of patients;
3. All radiographic units used for chest x rays have the exposure measured (with ± 50 percent accuracy) for a typical (23 cm thick) patient and this information be made available to the patient upon request;
4. Measurements of patient exposure always include a quantitative measure of image quality;
5. The use of chest radiography under the supervision of a non-radiologist be studied;
6. Methods be developed to reduce repeat examinations caused either by improper technique or irretrievable films; and
7. The role of hospital accreditation agencies in the reduction of patient exposure be investigated.

QUALITY ASSURANCE

Studies by the Bureau of Radiological Health and the State Radiation Control Officers have demonstrated that a major source of unnecessary x-ray exposure and poor quality radiographic images is improperly operating x-ray equipment.

We recommend:

1. All medical facilities with x-ray equipment have a radiographic quality assurance (QA) program. The QA program should monitor each of the physical parameters involved in the x-ray process in order to obtain good image quality at a reasonable exposure;
2. The NIOSH (pneumoconiosis) quality assurance program for chest radiography be studied for its applicability on a nationwide scale; and
3. The present coooperation between BRH and ACR in the establishment of quality assurance programs be continued and expanded.

OBSERVER PERFORMANCE

There presently is no agreed-upon method for correlating observer performance measurements with the various physical factors involved in producing a chest radiograph.

The need for an adequate observer performance testing procedure was discussed in the section on Imaging Centers. Numerous studies have established that a great deal of information is lost due to poor observer performance. There are many measurements which can be performed with present procedures.

We recommend:

1. Studies be performed on the utility of various techniques, e.g., stereo and lateral views, overpenetrated films;

2. Physical measurements and phantom studies be made on the various film/screen systems used in chest radiography (and the results of such studies be presented in an easily understood format);

3. Methods of improving observer accuracy, such as routine double-viewing, be investigated; and

4. Studies be performed to determine the best techniques (e.g., kVp, film systems) to detect different pathological conditions.

APPENDIX B

BIBLIOGRAPHY

A part of the contract for this symposium was a review of the scientific literature on chest radiography. The contents of this bibliography were primarily obtained from a Medline search covering the literature in the English language of the past 10 years. In addition, members of the symposium planning/advisory committee (listed below) gave many references.

The articles in this bibliography are subject coded, according to the general areas of:

E = Exposure
P = Perception or Viewer Performance
T = Technique

Not all articles were read and some of the subject areas designated were based on the title of the article.

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